University of Alberta

An Investigation of Repetitive Proprioceptive Reaching and Atypical Shoulder Muscle Activation in Multidirectional Shoulder Instability

by

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A thesis submitted to the Faculty of Graduate Studies and Research in partial fulfillment of the requirements for the degree of Doctor of Philosophy

in

Experimental Surgery

Department of Surgery

Edmonton, Alberta

Fall 2005

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Dedication

To Rhonda for her faith and patience, and to Samantha, Amanda, Nicholas and Dylan

Abstract

Multidirectional instability (MDI) of the shoulder is a disabling and painful condition that is difficult to diagnose and to treat. It has an equivocal atraumatic etiology that is frequently related to glenohumeral hyperlaxity and produces symptoms in the midrange positions of glenohumeral motion. These characteristics suggest that MDI is associated with compromised dynamic stability, which may be the result of deficient proprioception. The purpose of this research was to examine this relationship by investigating the dynamic proprioception of the upper limb and the muscle activation patterns of the shoulder. Several proprioceptive arm endpoint-repositioning tasks were used to compare the motor behaviour of subjects with MDI to subjects with healthy shoulders. This procedure also assessed the immediate effect of an elastic shoulder brace on proprioceptive reaching task performance in both groups. Electromyography (EMG) was used to assess the activity of the deltoid, pectoralis major, latissimus dorsi, infraspinatus and supraspinatus muscles in both groups during a series of controlled planar shoulder movements on an isokinetic dyanomometer. Two and three-way repeated measures analysis of variance (ANOVA) procedures were used to analyze the data statistically. When reaching to remembered proprioceptive targets, subjects with MDI were not able to reposition the endpoint of the limb as consistently as control subjects. Further, wearing an elastic shoulder brace had no immediate effect on the arm endpoint position and kinematic variables associated with proprioceptive reaching for subjects in either group. Subjects with MDI also demonstrated atypical patterns of activation in several different shoulder muscles, including the infraspinatus and supraspinatus muscles. The results of

these experiments are consistent with the hypothesis that multidirectional instability is associated with a proprioceptive deficiency of the shoulder.

Acknowledgement

This research received financial support from the Edmonton Orthopaedic Research Fund. Additional support was received from the Province of Alberta Graduate Fellowship program and a 75th Anniversary Scholarship Award from the Faculty of Medicine and Dentistry at the University of Alberta.

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1

General Introduction

1.1 Introduction

The human shoulder girdle is a multi-articulated functional unit whose distinct elements operate synchronously to produce unparalleled mobility. The primary articulation of the shoulder is the glenohumeral joint, which possesses the greatest range of motion of any joint in the human body (An & Friedman, 2000). However, the glenohumeral joint is also the most frequently dislocated (Moore *et al.*, 2004) due to its relatively poor osseous and capsuloligamentous stability (Nyland *et al.*, 1998). In the healthy shoulder, glenohumeral stability is achieved through a complex interaction of passive and dynamic factors. Passive (static) stability occurs as a consequence of the joint congruency between the head of the humerus and the glenoid fossa, the effect of the glenoid labrum (which increases the depth of the glenoid by as much as 50%), the extent of the glenoid version, negative intra-articular pressure (which provides a suction cup effect along with adhesion/cohesion at the articulation interface), and the tensile restraint provided by an integrated capsuloligamentous system (Bigliani *et al.*, 1996; Hurschler *et al.*, 2000; Jerosch *et al.*, 1996; Matsen III, 1994). Dynamic (active) stability occurs as a result of the voluntary and involuntary activation of muscles that control the motion of the humerus and scapula (Nyland *et al.*, 1998). A deficiency or failure of any one of these components can result in excessive intra-articular motion, which can precipitate further failure, thus stimulating a progressive cycle that may ultimately lead to joint instability (Lephart & Henry, 1996).

Glenohumeral instability (a.k.a. shoulder instability) is a disabling and painful condition that encompasses several different types of instability. Classifications for

shoulder instability are generally based on their etiology and the direction in which they occur (Gerber & Nyffeler, 2002). Shoulder instabilities that occur as a result of significant trauma can be identified through the resulting pathology (for example, a Bankart lesion is typically associated with anterior instability). While classification helps to delineate between different types of instability, the construction of a systematic and comprehensive taxonomy for glenohumeral instability is challenging because of the different criteria that can be applied. For example, shoulder instability can be classified as voluntary or involuntary, unilateral or bilateral, static or dynamic and acute or chronic. To the clinician, classification has indispensable practical value because prescription of the appropriate treatment depends upon an accurate diagnosis of the problem. Multidirectional instability (MDI), which involves instability of the glenohumeral joint in more than one direction, is a particularly complex form of glenohumeral instability that can be hard to classify and therefore diagnose (Mallon & Speer, 1995).

Multidirectional instability was first recognized as a unique and distinct clinical entity by Neer and Foster (1980), in which they prescribed a specific surgical procedure, the open inferior capsular shift, for its treatment. Multidirectional instability occurs in different age groups, both genders and different segments of the population, from sedentary individuals to elite athletes (Arendt, 1988). It involves recurrent subluxations and/or dislocations of the glenohumeral joint along with chronic or frequent episodes of pain. Symptoms are typically associated with the midrange positions of glenohumeral motion and often occur during activities of daily living (Beasley *et al.*, 2000; Schenk & Brems, 1998). The majority of patients with MDI have a loose and redundant joint capsule that produces an overly lax and hypermobile shoulder. In these cases, MDI may

occur bilaterally, although it is not uncommon for individuals to have bilateral hyperlaxity with unilateral MDI (Gerber & Nyffeler, 2002). Although MDI is frequently associated with shoulder and/or generalized joint laxity, it also occurs in a small minority of patients without glenohumeral hyperlaxity (Neer & Foster, 1980).

The discrimination between laxity and instability is further compounded by inconsistency in the etiology of capsuloligamentous laxity. Shoulder hyperlaxity that occurs synonymously with generalized ligamentous laxity is typically genetic in origin, while joint specific unilateral or bilateral shoulder laxity can occur as the result of repetitive trauma (An & Friedman, 2000). Because of the congenital component of glenohumeral hyperlaxity, MDI is typically regarded as an atraumatic condition, in reference to the fact that most cases of MDI occur without a single, major traumatic event (Kiss *et al.*, 2001; Neer & Foster, 1980). Consequently, traumatic lesions are typically absent although not in cases where MDI co-exists with concomitant unidirectional instability (Mallon & Speer, 1995). However, the classification of MDI as an atraumatic condition does not imply that it occurs in the absence of repetitive trauma. Several authors have suggested that minor repeated stresses of an everyday nature might be sufficient to induce a shoulder with excessive chronic laxity to become unstable (An & Friedman, 2000; Schenk & Brems, 1998). Reports indicate that a significant number of MDI patients are very active and athletic and have lifestyle-related factors that indicate the presence of repetitive stress (Neer & Foster, 1980). Consequently, based on etiology, there may be at least three different types of MDI: (1) purely atraumatic, which is directly related to congenital glenohumeral hyperlaxity, (2) repetitive microtraumatic, which precipitates or exacerbates existing glenohumeral laxity, and (3) macrotraumatic, which

requires the infliction of multiple traumatic events to produce a minimum of two concurrent unidirectional instabilities (An & Friedman, 2000; Brown *et al.*, 2000; Gerber & Nyffeler, 2002; Pagnani & Warren, 1994). The literature on MDI indicates that the majority of patients fall into the first two categories, which are essentially impervious to clinical differentiation (Mallon & Speer, 1995; Schenk & Brems, 1998). Therefore, the physician can only suspect the presence of repetitive trauma by obtaining comprehensive patient histories and descriptive screening information.

While the available evidence suggests that the pathogenesis of MDI is multifactorial, to the extent that MDI might be an overly inclusive classification representing more than one condition, its diagnosis is also complicated by the inconsistent relationship with glenohumeral laxity. This complication occurs because the presence of glenohumeral laxity does not necessarily imply instability (Brown *et al.*, 2000). An example is provided by inferior instability and its clinical manifestation, the sulcus sign, which are considered to be the diagnostic hallmarks of MDI (Neer & Foster, 1980; Mallon & Speer, 1995). Nevertheless, the presence of a sulcus sign does not necessarily indicate inferior instability because glenohumeral hyperlaxity with excessive inferior translation can occur without accompanying symptoms of instability (Brown *et al.*, 2000; Mallon & Speer, 1995). Consequently, the identification of shoulder instability, and MDI in particular, presents the physician with a considerable diagnostic challenge such that every effort must be made to identify the etiology of the instability through careful history taking, physical examination, and radiographic evaluation (An & Friedman, 2000). The difficulty associated with applying multiple criteria to the diagnosis of MDI is a problem to the extent that it has been suggested that a substantial

portion of the cases reported as MDI may actually be unidirectional instabilities with accompanying hyperlaxity, and that misuse of the term MDI has led to considerable confusion among orthopaedic surgeons (Gerber & Nyffeler, 2002).

The issues of classification, etiology and diagnosis underscore the importance of obtaining further information about the pathophysiology of MDI. Several hypotheses have been proposed that attempt to explain the causes of MDI, including a lack of coordinated activity in the rotator cuff muscles and the muscles responsible for maintaining a functionally stable glenoid platform (dysfunctional scapulohumeral rhythm), a general deficiency in glenohumeral neuromuscular control precipitated by deficient proprioception, and a deficiency of the collagenous material of the glenohumeral ligaments and joint capsule (Beasley *et al.*, 2000; Mallon & Speer, 1995; Schenk & Brems, 1998). Unfortunately, at the present time there is almost no evidence available to support or refute these hypotheses. This is not surprising given the inherent complexities of the shoulder girdle and the difficulties associated with using quantitative methods to assess glenohumeral mechanics and joint proprioceptive acuity. Nevertheless, the need for controlled experimental studies to investigate questions pertaining to the etiology and pathophysiology of MDI is significant if only as a starting point from which further investigations can proceed. Ultimately, an improved understanding of MDI should occur as a result of investigations that attempt to provide new information about the pathomechanics and neuromuscular control of glenohumeral instability.

1.2 Proprioception and joint stability

To facilitate limb movement and prevent joint injury it is necessary to constrain intra-articular motion and prevent abnormal joint excursion. The containment of joint motion is referred to as joint stability, which is essential for preserving the functional integrity of the articular system. Although the general concept of joint stability is well understood, the actual mechanics are exceedingly complex and vary according to the structure of each particular joint. As was presented in the previous section, joint stability occurs as the result of multiple factors, which can be grouped into passive and dynamic categories. Dynamic stability is particularly important because skeletal muscles produce tension in response to neural activation, thereby creating a complex system of muscle torques that can initiate, control, limit and prevent joint motion. Consequently, dynamic stability is an adaptive mechanism capable of responding to a wide range of joint positions and loading conditions.

While it is conceptually useful to organize the factors that contribute to joint stability into passive and dynamic categories, this distinction is essentially an artificial and unrealistic depiction of functional arthrology (Nyland *et al.*, 1998). Rather than being mutually exclusive, these components operate as a reciprocating and structurally integrated system to prevent unwanted and/or excessive joint motion. Such integration can be found in the glenohumeral joint, in which portions of the rotator cuff tendons adhere and integrate with the glenohumeral ligaments and joint capsule. Consequently, tension developed through rotator cuff activation directly affects capsuloligamentous tension and orientation, and *vice versa*, in that glenohumeral ligament tension has the

potential to modulate muscle activation via direct and indirect reflex pathways (Diederichsen *et al.*, 2002; Nyland *et al.* 1998).

The somatosensory system also provides an important link between the passive and dynamic components of joint stability. Muscles, tendons, ligaments and skin are innervated with specialized sensory receptors (mechanoreceptors) that provide information from which the sensations of position, movement, tension, balance and effort are derived (Proske, 2005). Collectively, the terms proprioception and/or kinaesthesia are used to refer to one or more of these sensations, although there is some discrepancy with respect to the inclusiveness of the definitions of these terms. The term kinaesthesia is used synonymously with dynamic proprioception to refer to the sensations elicited as a consequence of joint motion (motion sense). In contrast, static proprioception (position sense) refers to the conscious perception of the orientation of the different body parts to one another (Jerosch & Prymka, 1996a). The more general term "proprioception" is typically used to refer to both static and dynamic proprioception (Nyland *et al.*, 1998), although this too is often used synonymously with the term kinaesthesia (Proske, 2005). While the sensory information received from capsuloligamentous, cutaneous and musculotendinous mechanoreceptors is known as proprioceptive input, other types of sensory information, namely visual and vestibular input, in addition to information provided by central motor areas, also contribute to and influence the sensation of proprioception. Figure 1.1 presents a comprehensive schematic of the various factors influencing proprioception (Jerosch & Prymka, 1996a).

Numerous studies have demonstrated the importance of proprioceptive input to the neural control of movement of which joint stability is an important part. Proprioception is

essential for the coordination of multijoint movement sequences (Bevan *et al.*, 1994; Cordo *et al.*, 1994; Ghez *et al.*, 1990; Ghez & Sainburg, 1995), for regulating the amplitude and timing of antagonistic muscle activation for the purpose of limb segment deceleration (Forget & Lamarre, 1987), and for regulating the contraction of distal segment muscles to control proximally-produced joint interaction torques (Gribble & Ostry, 1999; Sainburg *et al.*, 1993; Smith & Zernicke, 1987). Studies of the relationship between proprioception and joint stability acknowledge the complementary role of sensorimotor systems and passive joint structures to the maintenance of functional joint stability (Beard *et al.*, 2000; Iwasa *et al.*, 2000; Jerosch & Prymka, 1996a). Proprioception provides a mechanism by which the neuromuscular control system operates as a kind of injury prophylactic to maintain the functional integrity and stability of the joint (Beard *et al.*, 2000; Iwasa *et al.*, 2000; Jerosch & Prymka, 1996a). Muscle stiffness, muscle response latencies, and passive and dynamic (active) joint-repositioning tests have all been used to attempt to assess proprioceptive function indirectly (Beard *et al.*, 2000; Brindle *et al.*, 1999).

To quantify joint proprioceptive acuity, the most common experimental approach has been to perform passive joint repositioning tests or measure the threshold to detection of passive motion. However, Grob *et al.* (2002) recently demonstrated that passive test procedures produce unreliable results and that they should not be used to infer proprioceptive ability. Studies have investigated proprioception and its relationship to joint function as it applies to aging (Petrella *et al.*, 1997; Skinner *et al.*, 1984), osteoarthritis (Sharma *et al.*, 1997), ankle injury (Richie Jr., 2001), knee injury (Carter *et al.*, 1997; Pap *et al.*, 1999), traumatic anterior shoulder instability (Lephart *et al.*, 1994;

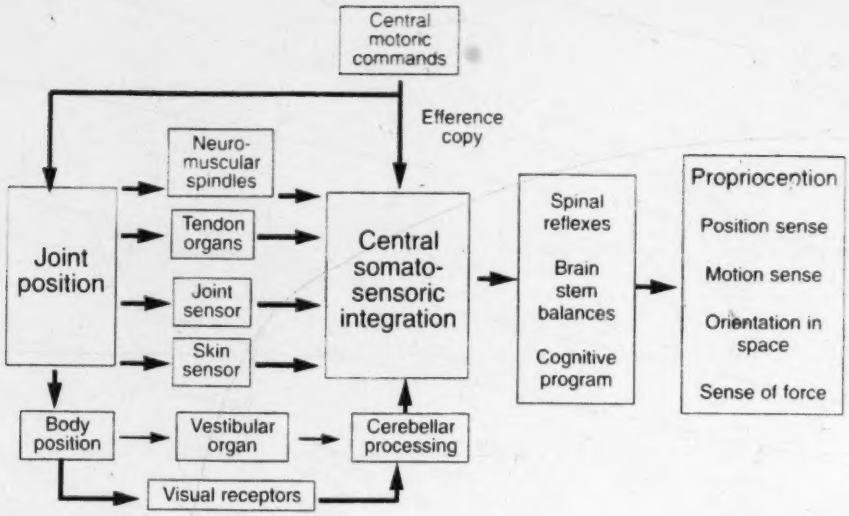

Figure 1.1: *Factors that influence proprioception. Reproduced from Knee Surgery, Sports Traumatology, Arthroscopy, 4, 1996, p. 172, Proprioception and joint stability, J. Jerosch & M. Prymka, Figure 1. © 1996 by Springer International, Inc. With kind permission of Springer Science and Business Media.*

Zuckerman *et al.*, 2003) and elastic bracing (McNair *et al.*, 1996; Perlau *et al.*, 1995; Risberg *et al.*, 1999; Wu *et al.*, 2001). Numerous studies have focused on the knee and issues relating to the injury, reconstruction and rehabilitation of the anterior cruciate ligament (ACL). These studies have indicated that patients who experience an ACL injury possess deficient knee proprioception compared to participants with healthy knees (Jerosch & Prymka, 1996a). The pathogenesis of deficient joint proprioception is not well understood and a particularly important question concerns whether proprioceptive deficiency precedes or follows the infliction of joint injury. If deficient proprioception is

a pre-existing condition, it may predispose an individual to joint instability and/or injury. Conversely, it may occur as a consequence of acute or repetitive trauma to the capsuloligamentous structures of a joint.

The correlation between joint instability and deficient proprioception was first described by Kennedy *et al.* (1982), in which they proposed a cycle of joint instability that occurred as a result of damage to the mechanoreceptor-containing structures of the knee. More recently, Lephart and Henry (1996) have formalized a conceptual model called the Functional Stability Paradigm to describe the pathogenetic process of joint instability (Figure 1.2). In this model, damage (either through acute or repetitive trauma) to the capsuloligamentous structures of a joint precipitates a loss of mechanical stability, presumably through the rupture, partial separation and/or plastic deformation of the ligaments. This loss of mechanical stability predisposes the joint to experience further excessive unphysiologic joint motion, thereby exacerbating ligamentous damage and joint laxity. The model also envisions a parallel process in which the structural ligamentous injury either precipitates or is accompanied by neurological damage. In this process, damage to joint mechanoreceptors results in a loss of proprioceptive input, with subsequent deleterious effects on the neuromuscular control required to maintain joint stability. While the scope of the functional stability paradigm is limited to the effects of deficient proprioception on joint stability, it is also reasonable to propose that damage to joint proprioceptive afferents should also affect the neuromuscular control of the entire linked system of which that joint is a part. Therefore, if the functional stability paradigm is correct, decreased neuromuscular control in the affected joint should produce motor performance deficits associated with the performance

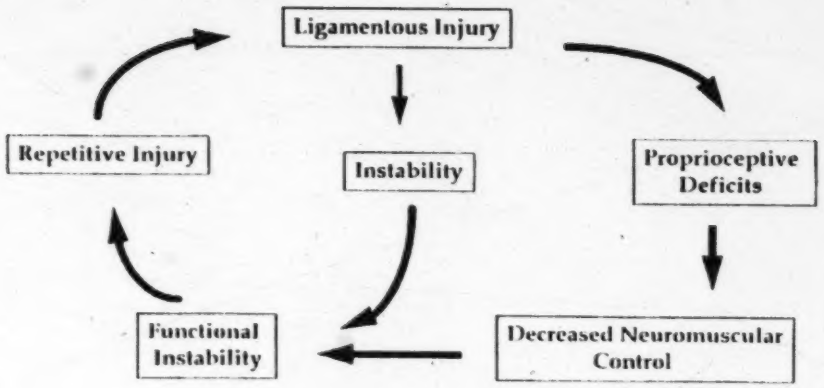

Figure 1.2: *Functional stability paradigm depicting the influence of mechanical instability and proprioceptive deficits on functional stability, which predisposes the joint to repetitive injury. From S.M. Lephart, and T.J. Henry, 1996, The physiological basis for open and closed kinetic chain rehabilitation for the upper extremity, Journal of Sport Rehabilitation 5(1): 78, figure 1. © 1996 by Human Kinetics Publishers, Inc. Reprinted with permission from Human Kinetics (Champaign, IL).*

of functional multijoint limb movement tasks.

With respect to the glenohumeral joint, only a small number of studies have investigated the proposed relationship between deficient proprioception and shoulder instability (Blasier *et al.*, 1994; Lephart *et al.*, 1994; Warner *et al.*, 1996; Zuckerman *et al.*, 2003). These studies have employed passive joint repositioning tests to assess proprioceptive function in patients with recurrent anterior instability and in all cases the joint position sense of subjects with instability was found to be inferior to that of the healthy participants. Given its relationship to glenohumeral hyperlaxity and its atraumatic

etiology, it is reasonable to suspect that deficient proprioception is one of several possible factors related to the pathophysiology of MDI. Several authors have suggested this hypothesis (Mallon & Speer, 1995; Schenk & Brems, 1998) and it is surprising that no attempt has been made to investigate the presence of upper limb proprioceptive deficits in subjects with MDI, given the importance of neuromuscular control to glenohumeral stability (Nyland *et al.*, 1998) as well as the importance of proprioceptive input to the sensorimotor control of movement, of which joint stability is an integral part. This is particularly remarkable given the interest in the contribution of proprioceptive feedback mechanisms to functional joint stability in the knee and to injuries of the ACL (Beard *et al.*, 2000; Beynnon *et al.*, 1999; Birmingham *et al.*, 1998; Birmingham *et al.*, 2000; Iwasa *et al.*, 2000; Jerosch & Prymka, 1996b).

1.3 Aim of the research

The purpose of this research was to investigate two hypotheses that have been proposed in the literature to explain the etiology and pathophysiology of MDI. One is that MDI is associated with a proprioceptive deficiency in the symptomatic shoulder. A specific mechanism for this hypothesis has not been proposed although the Functional Stability Paradigm is applicable in that it describes a general model for the pathogenesis of joint instability (Lephart & Henry, 1996). The second hypothesis, which is potentially related to the first, is that MDI is associated with dysfunctional activation of one or more of the rotator cuff muscles. Both of these hypotheses suggest a breakdown in the neuromuscular control mechanism that provides dynamic stability to the glenohumeral joint, which is

consistent with the observed symptoms (i.e., midrange instability) and the atraumatic etiology of the condition.

To test the hypothesis of deficient proprioception, a test was devised in which subjects relied on proprioception to perform several repetitive, three-dimensional, goal-directed reaching tasks. It was reasoned that a deficit in the peripheral afferent input from the shoulder should have a negative effect on the sensorimotor control of the arm under conditions in which vision was removed. Although a number of mechanisms are possible, it was conceived that insufficient sensory reafference would compromise the ability of an inverse dynamics model to generate the appropriate feedforward motor commands to precisely control the positioning of the limb in three-dimensional space. Given that proprioceptive acuity must be inferred from performance and cannot be measured directly, an active limb-repositioning test was devised so the results would be generalizable to the actual conditions in which MDI occurs, namely instability during volitional muscle activation in positions not at the extremes of motion. It was decided not to employ passive test procedures for a number of different reasons, but most importantly because passive tests have been shown to be unreliable (Grob *et al.*, 2002) and are not representative of how a functioning neuromuscular control system operates (Beynnon *et al.*, 2002).

In relation to the hypothesis of deficient proprioception, it was also thought that an elastic shoulder orthosis might be able to ameliorate an existing proprioceptive deficiency by enhancing joint cutaneous feedback, which has been proposed as a potential mechanism for how elastic braces influence proprioceptive acuity and joint stability. Consequently, the experimental methodology described above was also used to test

subjects with MDI in braced and non-braced conditions, to determine if the brace was able to provide any immediate benefit when using proprioception to control the positioning of the arm when performing repetitive goal-directed tasks.

To test the hypothesis of dysfunctional rotator cuff activation, the activation patterns of the supraspinatus and infraspinatus muscles were investigated along with several other shoulder muscles when performing constrained planar shoulder movements on an isokinetic dynamometer. If the processes leading to the progression of joint instability as described by the Functional Stability Paradigm are correct, then atypical muscle activation patterns in the shoulder should be expected as a necessary consequence of a proprioceptive deficiency that precipitates a loss of neuromuscular control and the degradation of dynamic glenohumeral stability. Although there are several other potential mechanisms through which atypical shoulder muscle activation might occur in MDI, it was determined that this was an important hypothesis to test, that at the time had not been addressed in the literature, and was one in which positive results would provide support for the first part of the thesis.

1.4 Thesis contents

Chapter 2 presents brief reviews of the relevant literature pertaining to the major concepts presented in this paper format thesis.

The first paper (Chapter 3) investigates repetitive pointing movements to remembered proprioceptive targets in subjects with normal, healthy shoulders. The purpose of this study was to determine whether dynamic proprioception could be used to

modify the initial sensorimotor conditions associated with an active definition of a virtual proprioceptive target. This study extends previous work on reaching to proprioceptive targets because it used repetitive three-dimensional pointing tasks, rather than discrete tasks (Adamovich *et al.*, 1998; Adamovich *et al.*, 1999) or 2D tasks in which the arm was artificially supported against gravity (Brown *et al.*, 2003a; Brown *et al.*, 2003b), to assess reaching task performance. This study provides a basis for the methodology used to assess dynamic upper limb proprioception in subjects with MDI. Hand trajectory start and endpoint positions were quantified for each repetition and from these coordinates three-dimensional pointing errors relative to the target position and the initial and preceding movement repetitions, as well as changes in movement direction and extent, were determined. This paper was published in the journal *Human Movement Science* in 2005 (Barden *et al.*, 2005b).

Chapter 4 presents a paper in which the ability of subjects with MDI to use dynamic proprioception to return the hand repeatedly to a remembered proprioceptive target was assessed and compared to control subjects with normal, healthy shoulders. This paper was published in the journal *Clinical Orthopaedics and Related Research* in 2004 (Barden *et al.*, 2004).

In Chapter 5, the methodology used in Chapter 4 to assess dynamic upper limb proprioception was extended to incorporate the use of tangential hand velocity profiles to analyze a single component of the kinematics associated with proprioceptive reaching. Numerous studies have used tangential hand velocity profile analysis to identify various principles and invariant characteristics associated with the neural control of reaching movements (Gordon *et al.*, 1994; Lacquaniti *et al.*, 1982; Milner & Ijaz, 1990; Nagasaki,

1989; Soechting & Lacquaniti, 1981; Wolpert *et al.*, 1995). In this paper, the hand trajectory target and endpoint inter-repetition errors and velocity profile kinematics of subjects with MDI were compared to control subjects with normal, healthy shoulders for repetitive reaching movements to remembered proprioceptive targets.

Chapter 6 presents a paper in which the immediate effect of an elastic shoulder brace on the ability of subjects with MDI to use proprioception to perform repetitive, goal-directed reaching tasks was investigated. The effect of this orthosis on a group of healthy participants is also presented.

Chapter 7 presents a paper in which surface and intramuscular fine-wire electromyography (EMG) was used to compare the shoulder muscle activation patterns of subjects with MDI to control subjects with normal, healthy shoulders. Subjects performed repetitive shoulder abduction/adduction, flexion/extension and internal/external rotation movements on an isokinetic dynamometer in which the activity of the deltoid, infraspinatus, supraspinatus, latissimus dorsi, and pectoralis major muscles were recorded. Group differences in the amplitude, onset, termination and duration of the muscle activity were compared. This paper was published in the journal *Clinical Neurophysiology* in 2005 (Barden *et al.*, 2005a).

Chapter 8 provides a general discussion of the research conducted for this thesis. Conclusions based on the findings of the studies are presented as well as several recommendations for future research.

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2

Review of Literature

2.1 Multidirectional shoulder instability

In 1980, Neer and Foster (1980) published a seminal paper that identified the specific condition of multidirectional instability (MDI) of the shoulder. Prior to this publication, the diagnosis of MDI did not exist in the medical literature. Neer and Foster (1980) defined MDI as a unique and distinct clinical entity from other types of glenohumeral instability and prescribed a specific surgical procedure, the open inferior capsular shift, for its treatment. Since then numerous clinical studies and reviews have described the history, classification, diagnosis (clinical, radiographic and arthroscopic examination procedures), biomechanics, etiology, pathophysiology, and treatment modalities associated with MDI (An & Friedman, 2000; Arendt, 1988; Brown *et al.*, 2000; Kiss *et al.*, 2001; Lephart *et al.*, 2002; Mallon & Speer, 1995; Miniaci & McBirnie, 2003; Schenk & Brems, 1998). These studies and reviews have been instrumental in identifying several critical issues associated with the etiology, diagnosis and treatment of MDI.

One of these issues concerns the complex and multifactorial etiology of MDI. Most patients have an overly lax and hypermobile shoulder, which may or may not be associated with generalized ligamentous laxity (Gerber & Nyffeler, 2002). Laxity, however, is not necessarily synonymous with instability and individuals often have unilateral MDI with bilateral hyperlaxity. Multidirectional instability is considered to be an atraumatic condition because the majority of cases occur without a single, major traumatic event (Kiss *et al.*, 2001; Neer & Foster, 1980) and involve glenohumeral laxity, either developed congenitally or through repetitive trauma (An & Friedman, 2000; Schenk & Brems, 1998). However, although traumatic lesions are typically absent, they

can occur in certain cases where MDI co-exists with concomitant unidirectional instability (Mallon & Speer, 1995). Because etiology is one of the fundamental criteria for the classification of instability, this ambiguity makes the diagnosis of MDI a challenge for the physician.

A second critical issue, which is related to the first, is that the literature has recently begun to describe several different variants of MDI. Based on etiology, the following different types can be identified: (1) purely atraumatic, which is directly related to congenital glenohumeral hyperlaxity, (2) repetitive microtraumatic, which precipitates or exacerbates existing glenohumeral laxity, and (3) macrotraumatic, which requires the infliction of multiple traumatic events to produce a minimum of two concurrent unidirectional instabilities (An & Friedman, 2000; Brown *et al.*, 2000; Gerber & Nyffeler, 2002; Pagnani & Warren, 1994). However, the identification and description of different variants is further confounded because it depends on the system of classification used.

Classifications for shoulder instability are based on their etiology and the direction in which they occur (Gerber & Nyffeler, 2002). Traumatic shoulder instabilities are typically identified in relation to the resulting pathology, such as the occurrence of a Bankart lesion (a separation of the labrum from the anterior aspect of the glenoid rim), which is frequently associated with anterior instability. While classification helps to delineate between different types of instability, the construction of a systematic and comprehensive taxonomy for glenohumeral instability is challenging because of the different criteria that can be applied. For example, shoulder instability can be classified as voluntary or involuntary, unilateral or bilateral, static or dynamic and acute or chronic.

Multidirectional instability (MDI), which by definition requires instability in at least two directions, is a particularly complex form of glenohumeral instability that can be hard to classify, and is therefore difficult to diagnose (Mallon & Speer, 1995). The issues of etiology and classification are particularly important because they have a significant impact on the accuracy and consistency of the clinical data reported in the literature. For the physician to make informed and intelligent decisions about the prescription of the appropriate treatment, it is essential to have unambiguous information about the description and classification of MDI. It is equally important to be able to refer to accurate clinical data in the literature to evaluate the outcomes of different procedures. Consequently, several authors have reported that the diagnosis and treatment of MDI is plagued with difficulty and confusion (Gerber & Nyffeler, 2002; Lo *et al.*, 2004).

Studies have shown that recurrence for MDI is common and that success rates for surgery are lower than for other types of instability (Kiss *et al.*, 2001). Assessing the effectiveness of different treatments is problematic because not all types of MDI respond to treatment in the same way (Kiss *et al.*, 2001). Treatment alternatives for MDI are typically grouped into operative and non-operative categories and, of these, it is generally accepted that non-operative procedures should be prescribed first. These methods typically include a program of regimented physical therapy aimed at strengthening the rotator cuff and scapular stabilizing muscles of the shoulder (An & Friedman, 2000; Arendt, 1988; Mallon & Speer, 1995; Kiss *et al.*, 2001; Schenk & Brems, 1998). Studies indicate that success rates for conservative treatment are high and that improved neuromuscular control can compensate, to a large degree, for deficient passive anatomical restraint mechanisms (Burkhead & Rockwood, 1992). Nevertheless, it is

reported that approximately 20% of patients with MDI do not respond well to conservative treatment (An & Friedman, 2000). In these cases, several different surgical procedures are available.

Given that the hallmark of MDI is capsular redundancy, the open inferior capsular shift as described by Neer and Foster (1980) has traditionally been the most favoured procedure (Brown *et al.*, 2000; Lo *et al.*, 2004). However, arthroscopic stabilization techniques such as thermal capsulorrhaphy, which uses thermal energy to shrink the joint capsule, have recently gained in popularity (Brown *et al.*, 2000; Lephart *et al.*, 2002). Several studies have suggested that these methods compare favourably with those of open surgery (Duncan & Savoie, 1993; McIntyre *et al.*, 1997) while others have suggested that they do not (Lo *et al.*, 2004; Miniaci & McBirnie, 2003). In a recent study of 19 patients with MDI, Miniaci & McBirnie (2003) found that thermal capsular shrinkage produced substantial failure rates and was associated with a range of postoperative complications including stiffness and a variety of neurological symptoms. Consequently, it was suggested that further investigation of the long-term outcomes of arthroscopic techniques should be conducted before they are used to replace the standard procedures of open surgery.

2.2 Peripheral and central mechanisms associated with joint proprioception and kinaesthesia

The terms proprioception and kinaesthesia, which are frequently used interchangeably to describe the sensations of joint motion and position (Gandevia, 1996), also describe a

number of other sensations associated with the generation and/or experience of movement. These terms encompass sensations that were originally proposed by Sherrington (proprioception) and Bastian (kinaesthesia) and arise as a consequence of the transduction of stimuli from specialized sensory receptors known as mechanoreceptors, which can be found in muscles, tendons, ligaments and skin (McCloskey, 1978). In addition to peripheral afferent input, these sensations are also affected by centrally-produced motor commands and the interactions of efferent and afferent signals (Gandevia, 1996). In general, the term proprioception is used to describe a number of sensations that arise from multiple stimuli and the current view is that the afferent and efferent signals have no unique specific role; each contributes to proprioception, volitional motor control and the reflexive regulation of limb movement and joint stability (Gandevia, 1996; Proske, 2005).

Joint mechanoreceptors transform mechanical stimuli (e.g. tension) into afferent input and can be classified into 4 types based on their activation characteristics (Nyland *et al.*, 1998). Type 1 (Ruffini endings) are low-threshold, slowly-adapting position sensors found within capsuloligamentous tissues and skin. Type 2 (Pacini, Vater-Pacini and Golgi-Mazonni) are low-threshold, rapidly adapting movement/velocity sensors found mainly within capsuloligamentous tissues, muscle fascia and skin. Type 3 (Golgi) are high-threshold, slowly-adapting tension sensors that contribute to the sensation of tension and provide assistance to muscle spindles in the calibration of position sense. Type 4 are most common in fibrous joint capsule layers and are generally classified as pain receptors (nociceptors) although their presence within intramuscular connective tissue is also believed to serve a joint proprioceptive function (Cleland & Rymer, 1990;

1993). Studies have shown that mechanoreceptor distribution is non-uniform within a specific joint as well as between different joints (Gandevia, 1996; Prochazka, 1996). Muscle spindles, which are currently believed to provide the predominant signal for position sense (Walsh *et al.*, 2004; Proske, 2005), have traditionally been classified into two basic types based on their afferent projections. Primary, large afferent axons project from dynamic bag_1 and nuclear chain fibres, whereas secondary, smaller afferent axons project from nuclear chain and static bag_2 fibres. It has also recently been reported that mammals have axons (skeletofusimotor) that innervate both intrafusal and extrafusal muscle fibres (Nyland *et al.*, 1998).

Several important properties of mechanoreceptors in relation to their function as position sensors have emerged, namely that muscle contraction aids joint afferent stimulation and increases the angular range over which joint afferents respond to passive movement (Gandevia, 1996; Goodwin *et al.*, 1972; McCloskey, 1978). Increases in joint afferent discharge have been found to be proportional to increases in joint stress as the joint moves into positions that approach the limits of its range of motion (Prochazka, 1996). Other studies have shown that angular thresholds for joint position sensitivity are less for proximal joints than for distal joints, indicating that proximal joints have greater proprioceptive acuity (Gandevia, 1996). This concept is supported by the work of Hall & McCloskey (1983) who demonstrated that the contribution of proprioception to the control of the upper limb is most evident at the most proximal joint, the shoulder, as opposed to the elbow or wrist joints. It has also been demonstrated that the capacity of cutaneous afferents to generate perceived signals of joint position and movement is less well developed than those of the other proprioceptive afferents and that cutaneous

mechanoreceptors make a greater contribution to proprioception at distal joints, such as the interphalangeal joints of the finger, than at proximal joints such as the shoulder (Voight *et al.*, 1996). With respect to peripheral afferent input, this suggests that proprioception at the shoulder joint depends primarily on musculotendinous and joint mechanoreceptor feedback.

In addition to joint motion and position, the terms proprioception and kinaesthesia also describe a group of sensations related to the production of muscle force (Gandevia, 1996). In general, the sensations derived from proprioceptors are categorized to include the kinaesthetic sense (i.e., the sense of joint position and movement), the sense of tension, the sense of balance and the sense of effort or heaviness (Proske, 2005). The current view with respect to the primary mechanisms responsible for the production of these sensations is that muscle spindles and joint afferents are responsible for the sense of joint position and movement, Golgi tendon organs are responsible for eliciting the sensation of tension (force), the vestibular system contributes to the sense of balance, and the sense of effort is generated within the CNS (Proske, 2005). However, the ability of the CNS to estimate the generation of muscle forces and the heaviness of objects appears to be partially related to the sense of effort. Studies have suggested that both the sense of effort and the sense of force and heaviness depend on peripheral afferent input as well as on centrally-derived signals related to the production of motor commands (Gandevia, 2001; McCloskey, 1978; Walsh *et al.*, 2004). A number of experiments have demonstrated changes in perceived force when the actual force remained unchanged and provide the basis for the suggestion that the CNS uses more than just the sensory signals

derived from muscle tension to estimate force and heaviness (Saxton *et al.*, 1995; Walsh *et al.*, 2004).

Although much remains to be determined about the specific central mechanisms associated with the generation of the sense of effort (Proske, 2005; Walsh *et al.*, 2004), it is well accepted that this sensation is derived from a corollary discharge of the motor command (Matthews, 1988) and that it occurs centrally as the result of interactions between the corollary discharge and the peripheral afferent input (Proske, 2005). The current view is that the sense of effort arises as a consequence of the interaction between the sensory cortex and the motor cortex (Gandevia, 1996; McCloskey *et al.*, 1983; Prochazka, 1996) although which specific part (or parts) of the brain are responsible for producing this sensation remains to be determined. Corollary discharges are also thought to play an important role in the ability of the CNS to extract the information provided by muscle spindles about the change in muscle length (see Fig. 2.1). Studies that have used muscle vibration to induce illusory movement effects have shown that voluntary contractions diminish or eliminate the illusory sensations (Goodwin *et al.*, 1972; Gandevia, 1996; McCloskey, 1978), suggesting that a central subtraction process occurs in which the fusimotor-evoked spindle activity is separated from the length-related discharge (McCloskey *et al.*, 1983).

Walsh *et al.* (2004) have also shown that the effort required to maintain the position of the arm against the force of gravity also provides information about the position of the limb in space. The idea that the sense of effort contributes to the determination of joint position is not new and has been suggested in several earlier studies (Soechting, 1982; Worringham & Stelmach, 1985). A working hypothesis with respect to how the sense of

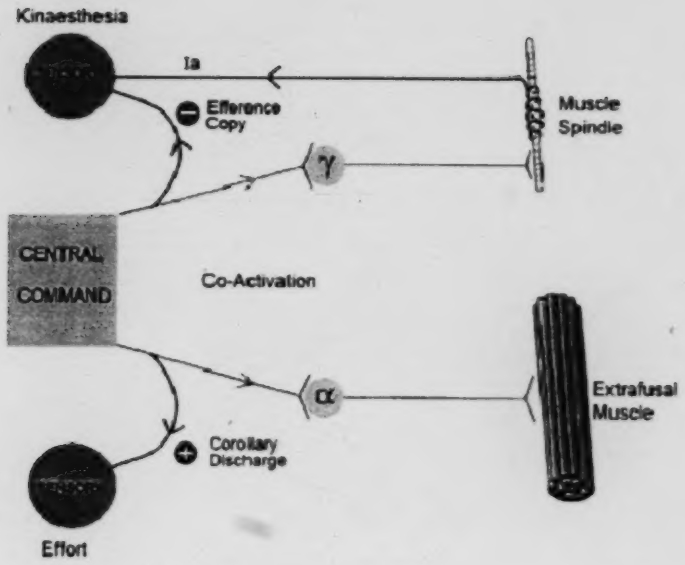

Figure 2.1: *Diagrammatic representation of postulated central feedback processes accompanying motor output. Motor command signals are directed to the spinal motor neurons (α) to initiate muscle contraction. The command signal is fed back as a corollary discharge to sensory areas to generate a sense of effort. Accompanying the command to spinal motor neurons is a signal to fusimotor neurons (γ) as a result of coactivation. A copy of the signal to fusimotor neurons, the efference copy, is fed back to sensory areas where this is subtracted from the total signal fed back by muscle spindles. The total signal comprises activity as a result of the change in muscle length, the exafference, plus the activity from the fusimotor activation, the reafference. Reproduced from* <u>*What is the role of muscle receptors in proprioception*</u>*?, U. Proske, Muscle & Nerve, Copyright © (1996 John Wiley & Sons). Reprinted with permission of John Wiley & Sons, Inc.*

effort might be related to the estimation of limb and/or joint position is that when the arm is moved to a given spatial location, movement-related information is provided via muscle spindles with additional contributions from the cutaneous and related joint afferents. Once the arm is in position, the effort required to keep it there provides additional detailed positional information (Walsh *et al.*, 2004). This supports the results of several other studies that have suggested that position information may have a different origin from movement information (Brown *et al.*, 2003; Sittig *et al.*, 1985).

In the glenohumeral joint, a limited number of morphological studies have documented the existence of mechanoreceptors within the joint capsule as well as in the glenohumeral, coracohumeral and coracoacromial ligaments (Gohlke *et al.*, 1998; Guanche *et al.*, 1999; Morisawa *et al.*, 1996; 1998; Vangsness, 1995). Various animal studies have also discovered Ruffini endings and Pacinian corpuscles in the shoulder structures of rabbits (Minaki *et al.*, 1999), pigeons (Halata & Munger, 1980), bats (Malinovsky & Novotny, 1980) and mice (Backenkohler *et al.*, 1996). In investigating the mechanosensitive capability of the glenohumeral joint in rabbits, Minaki *et al.* (1999) found that 53% of the mechanoreceptors were nociceptive and were situated in the rotator cuff, suggesting that the glenohumeral joint is particularly sensitive to noxious stimuli such as excessive joint motion. In addition, a greater number of mechanoreceptors were found in the supraspinatus muscle than in the infraspinatus.

The human glenohumeral joint is innervated by the suprascapular, axillary, lateral pectoral and subscapular nerves (Diederichsen *et al.*, 2002) and musculotendinous mechanoreceptors have been found in the insertions of the supraspinatus, infraspinatus and pectoralis minor muscles as well as in the origins of the biceps brachii, triceps

brachii, coracobrachialis, and deltoid muscles (Nyland *et al.*, 1998). Golgi tendon organ receptors in the supraspinatus and infraspinatus muscles have been found to occur in highly specialized muscle-tendon transition zones (Nyland *et al.*, 1998). Other studies have identified Ruffini endings, Pacinian corpuscles, free nerve endings and Golgi mechanoreceptors in the capsuloligamentous structures of the glenohumeral joint in addition to free nerve endings in the surrounding tissue (Guanche *et al.*, 1999; Hashimoto *et al.*, 1994; Vangsness *et al.*, 1995). In the human glenohumeral joint, Gohlke *et al.* (1998) reported Ruffini endings in the coracoacromial ligament and rotator cuff as well as Pacinian corpuscles in the joint capsule. A summary of the findings pertaining to the identification of mechanoreceptors in the periarticular structures of the glenohumeral joint is presented in Table 2.1.

Using a more functional approach, Jerosch *et al.* (1997) electrically stimulated the glenohumeral joint capsule of patients intraoperatively and found that the stimulation elicited muscle activity in all of the stabilizing muscles. The elicited activity possessed a relatively long latency period suggesting that it was an indirect reflex. In one of the most important studies of proprioception in the glenohumeral joint, Jerosch *et al.* (1993) also found that glenohumeral translation increased significantly following the administration of lidocaine (an anaesthetic) to the joint capsule and glenohumeral ligaments. Glenohumeral translation was assessed using anterior and posterior drawer tests and was measured using an established ultrasonographic procedure. In addition to demonstrating that the glenohumeral joint has proprioceptive capability, this finding also suggests that the increased translation occurred because the physiologic feedback mechanism between the joint capsule and the stabilizing musculature of the shoulder was disrupted.

Table 2.1: Periarticular mechanoreceptors of the glenohumeral joint in humans. Reprinted from Journal of Electromyography and Kinesiology (12), L. Diederichsen, M. Krogsgaard, M. Voigt, & P. Dyhre-Poulsen, Shoulder reflexes, p.186, Copyright (2002), with permission from Elsevier.

Authors, Method	Location of receptors	Types of receptors	Size of receptors (μm)	Total No. of receptors	Diameter of afferent fibers (μm)
Terade et al. [54] (1992): modified gold chloride Technique	Subacromial bursa	Free nerve endings and 4 types (types not reported)	Not reported	Not reported	Not reported
Yoshida et al. [62] 1992): immunohistochemistry	Subacromial bursa	Type 4 (Bipolar, pain nerves)	Not reported	Not reported	Not reported
Bresch et al. [4] (1994): Triton X100/ gold chloride staining technique	Middle and inferior glenohumeral ligaments	4 types (Freeman and Wyke): Type I–IV	Not reported	Not reported	Not reported
Robinson et al. [45] (1995): serial sections, modified gold chloride technique	Glenohumeral capsule, labrum and ligaments (inferior and superior)	4 types (Freeman and Wyke): Type I–IV	Not reported	Not reported	Not reported
Vangsness et al. [55] (1995): serial sections, modified gold chloride technique	Shoulder ligaments', glenoid labrum, subacromial bursa	4 types:	Not reported	Not reported	Not reported
		free nerve endings Ruffini Golgi tendon organs Pacini			
Mochizuki et al. [40] (1996): serial sections, modified gold chloride technique	Rotator cuff	4 types:	Not reported	Not reported	Not reported
	Subacromial bursa Coracoacromial ligament	free nerve endings Ruffini Golgi tendon organs Pacini			
Mochizuki et al. [41] (1998): serial sections, modified gold chloride technique	Coracoacromial ligament	4 types:			
		free nerve endings	100–500	9.2±6.3	
		Ruffini	200–300	3.8±4.9	
		Golgi tendon organ	Not reported	1.8±2.2	
		Pacini	100–200×50–100	3.5±3.6	
		Non-typical Pacini	100–200×50–100	3.8±2.5	
		Non-typical Ruffini	<200	3.8±2.0	
		Unclassifiable nerve endings	200–400	4.3±1.9	
Gohlke et al. [12] (1998): serial sections, immunocytochemistry	Glenohumeral capsule	4 types:			
	Rotator cuff	free nerve endings	Not reported50–440×100–700	Not reported	3–4
	Coracoacromial ligament	Ruffini	Not reported	61	6–9
		Golgi tendon organs	180	4	14–16
		Pacini		26	6
Guanche et al. [14] (1999): serial sections, modified gold chloride technique	Glenohumeral capsule, labrum and ligaments (inferior and middle)	4 types:	Not reported	Not reported	Not reported
		free nerve endings Ruffini Golgi tendon organs Pacini			

The results of this study were remarkably consistent with the clinical manifestations of MDI and provide support for the theory that deficient proprioception is a factor related to the development of this condition.

In a similar study, Voigt *et al.* (1998) found that non-noxious electrical stimulation of the anterior/inferior part of the glenohumeral joint capsule in conscious human subjects elicited a strong general inhibition of the muscles surrounding the joint. As was the case with the study by Jerosch *et al.* (1997), the elicited reflexes also had long latency periods and no short latency (direct) excitatory responses were found. The results of Voigt *et al.* (1998) also suggested a potent reflex pathway in which proprioceptive feedback can modulate the muscle activity of the glenohumeral joint. Most recently, Diederichsen *et al.* (2004) demonstrated a reflex pathway from afferents in the coracoacromial ligament to eight different muscles in the shoulder. These results have provided further insight into the neuromuscular relationship between joint afferents and muscle activity in the shoulder and suggest that in addition to providing mechanical stability, the coracoacromial ligament might make a more significant contribution to the muscular control and dynamic stability of the shoulder than previously thought.

In general, neurohistological studies have demonstrated the presence of a number of different mechanoreceptors within the glenohumeral joint and in the periarticular structures surrounding the joint. Although still preliminary, the literature has also presented evidence which supports the existence of an integrated and coherent mechanism between the proprioceptive afferents and the shoulder girdle musculature, including the rotator cuff, which has the potential to make a significant contribution to the control and stability of the joint.

2.3 Motor control systems

At the most basic level, the processes involved in the control of human movement can be conceptualized as being either open-loop or closed-loop control systems. In an open-loop control system, the motor output is planned in advance and control models of this type ignore the role of online sensory feedback and usually assume deterministic dynamics (Todorov, 2004). In contrast, closed-loop control systems incorporate feedback for the purpose of evaluation and modification and, in order to avoid sensory uncertainty and feedback delays, must contain some mechanism capable of prediction.

In reality, the CNS takes advantage of the benefits both processes provide. In the case of feedback control, the CNS can issue motor commands (corrections) in response to sensory information that is received as movements proceed. However, the CNS would not be effective if it was only reactionary and an effective control strategy is to pre-program movements in advance. This pre-programming of movement is called feedforward control, and considerable evidence exists to suggest that reaching movements are generated through a process in which the CNS executes a set of commands that are programmed in advance of the movement (Scheidt *et al.*, 2000). For feedforward control to be effective, the CNS must be able to predict the behaviour of the system (i.e., the sensory consequences of the action) given the specific conditions of the task (Scheidt *et al.*, 2000; Wolpert & Ghahramani, 2000). The idea that our nervous system has the capacity to predict the consequences of our own actions has emerged as an important theoretical concept that underlies all aspects of sensorimotor control (Wolpert& Flanagan, 2001). Consequently, one of the requirements of this method of

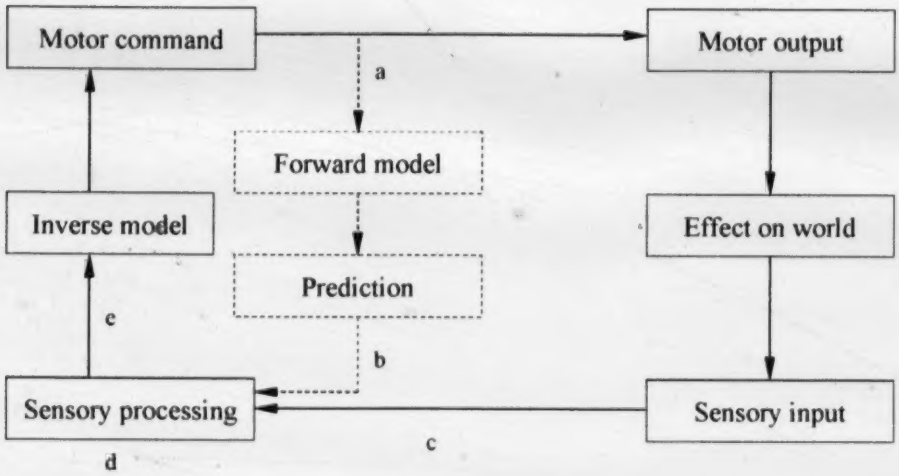

Figure 2.2: *Schematic representation of sensorimotor control with a forward model. The unbroken lines indicate the loop by which a motor command is translated into motor output, has some effect on the world, and creates sensory input that the system can process to generate the next motor command. The forward model is an internal loop (dashed lines) that takes the motor command and predicts the expected sensory input, which can be used to modulate the processing of the actual input (the reafference). The inverse model transforms the sensory error signal produced by the comparator (d) into motor terms (i.e., motor commands).* ***a*** *= efference copy;* ***b*** *= the prediction in sensory coordinates (expected reafference);* ***c*** *= the reafference;* ***d*** *= the estimator or comparator;* ***e*** *= the error signal (the difference between the sensory prediction & the reafference). Adapted from Webb (2004).*

control is that the feedforward controller must contain an internal model of how the system behaves. The basic elements of a feedforward control model are depicted in

Figure 2.2. An internal forward model is a predictor (or estimator) which can be defined as a neural process or procedure within the CNS that models the behaviour of the body and captures the forward or causal relationship between actions and their consequences (Wolpert & Flanagan, 2001). In order to do this, the internal forward model must have a way to transform motor commands into their sensory consequences. This transformation is governed by the physics of the environment, the inertial properties of the musculoskeletal system, the body's sensory receptors and the current state of the system. One of the important concepts related to prediction is that of efference copy which is often used synonymously with the term corollary discharge (Gandevia, 1996). This is essential because the forward model (predictor) must have input related to the action in order to be able to make a prediction about its outcome. This concept of being able to make predictions based on the motor command was originally proposed by Pukyne and Helmholtz and was simultaneously and independently more clearly formulated in the 1950's as "efference copy" by von Holst and Mittelstaedt, and as "corollary discharge" by Sperry (Webb, 2004). The important idea proposed by von Holst and Mittelstaedt was that the efference copy was designed to explicitly match the reafferent signal, whereas Sperry's idea of corollary discharge was that the motor output was fed back directly to central sensory areas to produce a perceived sensation in its own right (Gandevia, 1996; Proske, 2005). Evidence for the existence of efference copy comes from the studies of eye movements, in which the motor commands to the ocular muscles have been shown to contribute to the stabilization of the visual world and to help the CNS predict how much of an object's movement in retinal coordinates reflects the actual movement of the object and how much reflects the movement of the eye itself (Gandevia, 1996). Recently, it was

found that a pathway projecting from the superior colliculus to the frontal eye fields of the cerebral cortex, through the medio-dorsal thalamus, appears to convey corollary discharge information in non-human primates (Sommer & Wurtz, 2002).

The basic premise of motor prediction in the context of a reaching movement is that the CNS uses an internal forward model to predict how the arm will move given a certain set of conditions. Through a process that presumably models the complex dynamics of a multi-link, multiple degree of freedom system such as the arm, the outgoing motor commands can be transformed into predicted sensory consequences which can then be compared to the actual sensory feedback (the reafference). In this scheme, it is possible for a comparator to use the motor command to modulate the processing of the actual sensory input (Webb, 2004). This process produces an error signal that can be used as input to an inverse dynamics model. This process, which Flanagan *et al.* (2003) have referred to as "control" (as opposed to "prediction"), is necessary so that the desired consequences (the error signal) can be transformed into the appropriate motor commands.

Several advantages are conferred to the CNS through the use of internal models, one being that an internal forward model provides a mechanism with which the CNS is able to determine whether or not resulting actions are self-generated (Wolpert & Flanagan, 2001). In addition to being able to distinguish sensory reafference from the afferent input imposed by the external environment, this process enables the CNS to attenuate and/or enhance certain sensory stimuli (Proske, 2005). This idea is supported by studies that have found that the responses of neurons to sensory stimuli are altered depending on the context of the movement, such as the context of passive motion vs. actively maintaining a constant limb position (Scott, 2004). Another advantage of a

forward modelling control system is that problems related to the time dependency of sensory feedback are avoided. Sensory signals that convey information about the body's state (limb and/or joint position and velocity, for example) are subject to significant time delays due to receptor transduction, neural conduction and sensory processing (Wolpert & Flanagan, 2001). The use of a control scheme that incorporates internal models is also advantageous because the models can be trained. Studies indicate that internal models are not fixed entities (Hinder & Milner, 2003) but rather form the basis of an adaptive system that can learn from experience to produce new motor behaviours (Wolpert & Ghahramani, 2000).

Numerous studies have presented evidence in support of the concept that the CNS employs internal models to control movement (Flanagan & Wing, 1997; Lackner & Dizio, 1994; Lacquaniti *et al.*, 1992; Sainburg *et al.*, 1999; Shadmehr & Mussa-Ivaldi, 1994; Wolpert *et al.*, 1995). McIntyre *et al.* (2001) found that astronauts initiated catching movements earlier in a 0 g environment than in a 1 g environment, demonstrating that the CNS uses an internal model of gravity to estimate time to contact when catching a falling ball. Beer *et al.* (1999; 2000) have shown that the cortical damage that occurs as the result of a stroke or brain injury impairs the brain's ability to derive the appropriate information from the visual, vestibular and proprioceptive cues to generate an internal model of how the limb would behave under conditions of voluntary movement. Their findings were consistent with an impaired feedforward control mechanism that failed to compensate for the passive interaction torques that arise during multijoint movement (Gribble & Ostry, 1999; Sainburg *et al.*, 1995). It was concluded that this impaired control likely reflects a degradation in the internal representation of

limb dynamics that occurs either as a primary consequence of the brain injury or as a secondary symptom related to disuse. Ariff *et al.* (2002) have shown that the CNS relies on an internal estimate of arm dynamics to generate predictions about its position in space and Flanagan & Lolley (2001) have shown that internal forward models are capable of compensating for the effects of anisotropic limb inertia in reaching movements. In another study, Hinder & Milner (2003) demonstrated that subjects made errors in final position that were consistent with predictions based on the formation of a task specific internal dynamics model when the magnitude of a velocity dependent assistive torque was reduced.

With respect to the motor control of movement, other theories have suggested that motor programs specify, via the selection of a set of length-tension properties in agonist and antagonist muscles, an equilibrium point between sets of muscles to correctly position the arm in relation to the target (Abend *et al.*, 1982; Feldman, 1986; Polit & Bizzi, 1979). Others contend that the acquisition of a spatial target is accomplished via a series of successive approximations (Cordo & Bevan, 1992) or that movement amplitude, and not final position, is the primary kinematic variable controlled by the CNS (Bock & Eckmiller, 1986). Understanding the distributed nature of the motor control system is a challenging endeavour but it is becoming increasingly apparent that it is not possible to understand the neural mechanisms underpinning limb movement without a complete understanding of the underlying biomechanics (Lemon, 1999). That is to say, to understand movement it is necessary to understand both the system (plant) and the controller. To understand the plant it is necessary to have a complete description of both the active (skeletal muscle) and passive components of joint motion. Passive components

include the viscoelastic properties of passive tissue and the passive joint torques that arise from the effects of ligament, tendon and bone (Lemon, 1999).

With respect to the processes employed by the CNS to determine the appropriate strategy to perform a given task, another theory gaining in popularity is that of optimal control. The advantage of this theory is its ability to make elaborate and accurate predictions about the diverse behaviour of the motor control system in a variety of contexts and conditions (Todorov, 2004). In the theory of optimal control, feedback signals are optimized using a performance criterion that specifies the task objective, thus providing an important link between the sensory signals and the motor output of the system (Scott, 2004; Todorov, 2004). Given the task objective, optimal control theory asserts that the controller determines the specific details of the movement through a process in which the optimal control strategy or law is selected to achieve the best possible performance (Todorov, 2004). This control law is measured as the time integral of some cost function that the CNS attempts to minimize. Some examples of proposed cost functions are minimum jerk, torque change, energy or time. It is conceivable that the CNS may select different optimization criteria depending on the task objective, as it may not be possible (or optimal) to implement a single cost function for every possible situation.

Another important concept of optimal control theory is the idea that optimal feedback controllers obey a minimum intervention principle, such that movement corrections are only made for deviations that directly interfere with the attainment of the task objective. Consequently, the attraction of optimal control theory is that it provides a

parsimonious explanation for the inordinately complex behaviour that emerges from a highly distributed motor control system.

2.4 Proprioception and the neural control of movement

The peripheral afferent feedback received from the mechanoreceptors in muscles, tendons, ligaments, joints and skin plays an important part in the control of movement. Numerous studies have demonstrated that proprioceptive input is critical for the control of limb movement and body posture, for the specification of movement direction and extent and for the spatial steering and temporal coordination of multijoint movements (Bevan *et al.*, 1994; Brown *et al.*, 2003; Cordo *et al.*, 1994; Ghez *et al.*, 1990; Ghez & Sainburg, 1995; Hasan, 1992). The literature concerning the effects of proprioceptive afferents on motor output is extensive and comes from a variety of disciplines including orthopaedics, physical therapy, neurophysiology, neuroscience, biomechanics and motor control.

Studies have shown that proprioceptive input regulates rhythmic movement in a number of diverse species including locusts, turtles, stick insects and cats (Bässler, 1986; Conway *et al.*, 1987; Lennard, 1985; Wolf & Pearson, 1988), and recent evidence suggests this is also the case for walking in humans (Dietz, 2002). Other locomotor studies in cats (Gorassini *et al.*, 1994; Schomburg *et al.*, 1998) and in humans have demonstrated that lower limb motion modulates cutaneous reflexes with phase and task dependent properties (Dietz *et al.*, 1986; Dietz, 2002; Duysens *et al.*, 2002; Zehr *et al.*, 1997; Zehr & Haridas, 2003). These findings suggest that human interlimb coordination

in walking is mediated at the spinal level by propriospinal neurons which form part of specialized neural circuits in the caudal spinal cord known as central pattern generators (Dietz, 2002; Duysens *et al.*, 2002). There is also increasing evidence to support the idea that leg and arm movements for walking share similar neural control mechanisms, such that rhythmic arm movement is under the control of a common central pattern generator (CPG) that couples cervical and thoraco-lumbar propriospinal systems (Dietz, 2002; Zehr & Chua, 2000; Zehr & Haridas, 2003).

Studies that have taken more of a neuromechanical approach have shown that proprioceptive feedback is responsible for updating and adjusting the execution of learned motor patterns (Ghez *et al.*, 1990; Ghez & Sainburg, 1995; Gordon *et al.*, 1995; Karst & Hasan, 1991; Polit & Bizzi, 1979), for regulating the timing and amplitude of antagonistic muscle activation for the purpose of limb segment deceleration (Forget & Lamarre, 1987) and for monitoring changing limb dynamics when joint interaction torques increase to substantial levels (Sainburg *et al.*, 1993; Sainburg *et al.*, 1995; Smith & Zernicke, 1987). Studies of functionally deafferented patients with large-fibre sensory neuropathy have shown that, without proprioception, the CNS cannot adapt the amplitude and direction of applied forces to the actual mechanical properties of the limbs (Ghez *et al.*, 1995). These studies have identified the inherent problems associated with severe proprioceptive deficiency which include major deficits in movement control (Ghez *et al.*, 1995; Gordon *et al.*, 1994; Sainburg *et al.*, 1993; Sainburg *et al.*, 1995), increased movement variability (Marsden *et al.*, 1984; Sanes *et al.*, 1985), the inability to maintain stable hand postures without visual guidance (Sanes *et al.*, 1985) and a reduced capacity to detect and correct motion errors based on limb movement information following the

execution of steady-state motor commands (Ghez *et al.*, 1990; Gordon *et al.*, 1995; Marsden *et al.*, 1984; Sainburg *et al.*, 1993; Sanes *et al.*, 1985).

A related section of the literature is concerned with investigating the neural mechanisms responsible for the production of goal-directed reaching movements. These studies have found that constrained, point-to-point arm movements exhibit hand trajectories that are essentially straight (Abend *et al.*, 1982; Flash & Hogan, 1985; Gordon *et al.*, 1994; Morasso, 1981) with bell-shaped velocity profiles (Abend *et al.*, 1982; Atkeson & Hollerbach, 1985; Flash & Hogan, 1985; Gordon *et al.*, 1994) that are invariant with respect to the hand's initial and final position. In contrast, joint angular position and velocity profiles vary depending on the limb's workspace location (Morasso, 1981). Other studies have shown that pointing movements possess slightly curved trajectories, such that the degree of curvature depends on movement distance and direction (Atkeson & Hollerbach, 1985; Boessenkool *et al.*, 1998; Papaxanthis *et al.*, 1998; Wolpert *et al.*, 1995) and that the CNS uses a hand-centred coordinate system for the planning and execution of goal-directed reaching movements (Atkeson & Hollerbach, 1985; Brown *et al.*, 2003; Flash & Hogan, 1985; Kaminski & Gentile, 1989; Wolpert *et al.*, 1995). With respect to the use of internal models to control movement, a number of studies have presented evidence which suggests that motion-dependent proprioception modulates motor output by updating an internal model of the arm with movement specific limb dynamics information (Ghez *et al.*, 1990; Messier *et al.*, 2003; Sainburg *et al.*, 1995).

A coherent synthesis of the literature in this area is a challenging endeavour because different methodologies have been used to study the processes involved in reaching to

spatial targets (movements that are either single or multijoint, constrained or unconstrained, planar or multi-planar, continuous or discrete, with or without vision, slow or fast, etc.). The majority of studies have examined constrained, discrete (i.e., point-to-point) horizontal planar arm movements performed under a variety of different visual conditions, while a number of studies have begun to examine natural (multijoint), unconstrained arm movements performed in three-dimensional space (Adamovich *et al.*, 1998; Flanders & Soechting, 1990; Messier *et al.*, 2003; Milner & Ijaz, 1990; Soechting & Lacquaniti, 1981). With the introduction of commercially available three-dimensional motion capture technology in recent years, it is likely that an increasing number of studies will use natural, multi-joint arm movements to investigate the neural mechanisms associated with the planning and control of goal-directed reaching movements.

2.5 Proprioception and joint instability

The somatosensory system provides an important link between the passive and dynamic components of joint stability via peripheral afferent feedback. Proprioceptive input provides a mechanism by which the neuromuscular control system operates as a kind of injury prophylactic to maintain the functional integrity and stability of a joint (Beard *et al.*, 2000; Iwasa *et al.*, 2000; Jerosch & Prymka, 1996a). It has been suggested that joint instability is related to deficient proprioception in which damage to the mechanoreceptor-containing structures of a joint precipitates a chain of events that leads to joint instability (Kennedy *et al.*, 1982). Lephart and Henry (1996) have proposed a conceptual model called the Functional Stability Paradigm which suggests that damage (either through

acute or repetitive trauma) to the capsuloligamentous structures of a joint precipitates a loss of mechanical stability, presumably through the rupture, partial separation and/or plastic deformation of the ligaments. This loss of mechanical stability predisposes the joint to further excessive unphysiologic joint motion, thereby exacerbating ligamentous damage and joint laxity. This model also envisions a parallel pathogenetic process in which the structural ligamentous injury is compounded by neurological damage. In this parallel process, damage to the joint mechanoreceptors results in a loss of proprioceptive input, with subsequent effects on the neuromuscular control required to maintain dynamic joint stability.

In the last two decades, researchers have begun to investigate the concept of proprioceptive deficiency and its role in the etiology and pathophysiology of joint instability and injury. Studies have investigated proprioception and its relationship to joint function as it applies to aging (Petrella *et al.*, 1997; Skinner *et al.*, 1984), osteoarthritis (Sharma *et al.*, 1997), ankle injury (Richie Jr., 2001), knee injury (Carter *et al.*, 1997; Pap *et al.*, 1999) and shoulder instability (Lephart *et al.*, 1994; Zuckerman *et al.*, 2003). To assess proprioceptive function indirectly a number of methods have been used including the assessment of muscle stiffness and the determination of muscle response latencies (Beard *et al.*, 2000; Brindle *et al.*, 1999). However, the most common experimental approach has been to perform passive joint repositioning tests and/or to measure the threshold to detection of passive motion.

Several studies have suggested that proprioceptive deficiency is a contributing factor to joint instability at the knee (Beynnon *et al.*, 1999; Iwasa *et al.*, 2000) but a major focus of joint proprioceptive research has been on the injury, reconstruction and

rehabilitation of the anterior cruciate ligament (ACL). These studies have demonstrated that patients who have torn an ACL possess deficient knee proprioception compared to subjects with healthy knees (Beard *et al.*, 2000; Beynnon *et al.*, 1999; Birmingham *et al.*, 1998; Birmingham *et al.*, 2000; Iwasa *et al.*, 2000; Jerosch & Prymka, 1996a; Jerosch & Prymka, 1996b). With respect to the glenohumeral joint only a few studies have investigated joint proprioception as it applies to shoulder instability (Blasier *et al.*, 1994; Lephart *et al.*, 1994; Warner *et al.*, 1996; Zuckerman *et al.*, 2003) and none have investigated whether proprioceptive deficits are associated with MDI.

A potential difficulty with the literature on proprioception and joint instability concerns the reliability and generalizability of the passive test procedures that are typically used to assess joint proprioceptive acuity. The reliability of these procedures has been questioned because of disparate knee joint position sense values reported in the literature and it has recently been demonstrated that five different passive knee motion detection and joint reproduction tests produced five different sets of results in the same subjects, none of which were correlated with each other (Grob *et al.*, 2002). It has also been shown that passive tests for joint proprioception are non-functional and do not apply to the actual conditions in which the CNS coordinates muscle activity to produce intra-articular stability and inter-segmental motion (Beynnon *et al.*, 2002; Risberg *et al.*, 1999). Passive tests have also received criticism because they do not account for differences in muscle spindle sensitivity and may or may not control for cutaneous feedback from contact with the passive motion apparatus (Proske *et al.*, 2000). Studies that have used passive joint repositioning tests to assess proprioceptive function also contain a large number of methodological differences including the target angle used for joint

reproduction, the use of different directions and ranges of motion (for e.g., joint reproduction at different positions starting from either flexion or extension) and the use of different movement speeds for detecting passive motion. These methodological differences make the comparison of results between studies extremely difficult.

To improve the current knowledge about the relationship between proprioception and joint instability, it is necessary to conduct studies that use active test procedures to investigate the motor control of joint and/or limb movements. Studies also need to employ other methods, such as inverse dynamics analysis, to investigate the motor behaviour and joint kinetics associated with joint instability. Further research is necessary to understand the pathogenesis of deficient joint proprioception and to determine whether a proprioceptive deficiency precedes or follows the infliction of a joint injury. It is presently not known whether deficient proprioception is a pre-existing condition, which predisposes certain individuals to joint instability and/or injury, or if it occurs primarily in response to acute or repetitive trauma to the capsuloligamentous structures of a joint.

2.6 Proprioception and elastic bracing

Elastic braces are routinely used in a variety of clinical and sporting environments and are believed to augment joint stability by enhancing proprioceptive input (Birmingham *et al.*, 2000; Jerosch & Prymka, 1996b; McNair *et al.*, 1996; Perlau *et al.*, 1995; Wu *et al.*, 2001). This enhancement ostensibly occurs via the tight fitting compressive action of the brace which facilitates an increased discharge of cutaneous, joint and/or muscle spindle afferents (Perlau *et al.*, 1995; Wu *et al.*, 2001). Plenty of anecdotal and empirical

evidence exists concerning the benefits of elastic bracing but little is known about their effect on joint stability and the neuromuscular control of movement. Consequently, researchers have begun to investigate the relationship between elastic bracing and joint proprioception, particularly for the knee and ankle joints, and have relied on passive joint repositioning and motion detection tests to assess proprioceptive function.

Subjects with healthy knees (Birmingham *et al.*, 1998; McNair *et al.*, 1996; Perlau *et al.*, 1995), with ACL injury (Jerosch & Prymka, 1996b) and with ACL reconstructed knees (Lephart *et al.*, 1992; Wu *et al.*, 2001) have been found to demonstrate greater sensitivity on motion detection tests, and greater accuracy on passive-positioning tests, when wearing an elastic brace. In contrast, other studies have shown that wearing an elastic brace has no affect on knee joint proprioceptive acuity or stability (Beynnon *et al.*, 1999; Beynnon *et al.*, 2002; Risberg *et al.*, 1999). Given that contradictory evidence exists with respect to the effects of elastic bracing on joint proprioception at the knee, it is uncertain whether any neurological or functional mechanical benefits are to be gained from wearing an elastic brace.

It is likely that the conflicting results are at least partially related to the lack of standardized test procedures available to evaluate the effects of elastic bracing. In addition to the problems associated with passive joint repositioning and motion detection tests mentioned previously, other methodological issues exist, including the use of the contralateral uninvolved limb as a control limb. In these situations, negative findings would be expected if deficient joint proprioception were a pre-existing condition, such that the contralateral joint also possessed an underlying proprioceptive deficiency. Additional methodological discrepancies between studies include the assessment of

subjects with or without surgery, with or without parallel programs of physical therapy and the conduction of tests at different stages of the recovery period following injury (for example at 6 months, 12 months or 24 months).

Given the potential for elastic bracing to enhance joint proprioception in the ACL-injured or ACL-reconstructed knee and the popularity of elastic braces with athletes, trainers and physical therapists, it is surprising that their effects have not been investigated in relation to glenohumeral instability. This is particularly remarkable considering the importance of neuromuscular control mechanisms to the stability of the glenohumeral joint, and the implication that deficient neuromuscular control may be a factor in the pathophysiology of MDI. Consequently, further investigation into the relationship between elastic bracing and joint proprioception is necessary, both for the knee and for the shoulder. Future studies should investigate both the long- and short-term effects of elastic bracing on joint proprioception and neuromuscular control using a variety of different test procedures such as inverse dynamics analysis, which has been used to successfully quantify the effects of functional knee bracing on human locomotor movements (DeVita *et al.*, 1996; DeVita *et al.*, 1998; Ramsey *et al.*, 2003). Dynamic quantities such as joint moments and powers, obtained by non-invasive inverse dynamics analysis, should be able to differentiate between normal and abnormal movements with greater sensitivity than standard kinematic quantities, as has been shown in the case of patients with cerebellar ataxia (Riener & Straube, 1997).

2.7 Electromyography and shoulder instability

Electromyography (EMG) is an important tool in the study of human movement because it allows the muscle activity associated with the performance of a number of different tasks to be studied. In 1944, Inman *et al.* (1944) used quantitative EMG to present a comprehensive examination of the shoulder girdle. This study was significant not only because of what was determined about the function of shoulder muscles but because it scrutinized the muscle activity of a joint in motion. Consequently, it provided a revolutionary new approach to the investigation of muscle function. Since this time, this approach and the application of EMG has become commonplace in many different kinesiological, orthopaedic and ergonomic environments. Electromyography has been used to examine the amplitude and temporal characteristics of muscle activity in response to concentric or eccentric loading conditions (Öberg *et al.*, 1990), to determine the relationship between muscle activation and force production (Ringelberg, 1985), to provide biofeedback to patients during clinical rehabilitation (Pande *et al.*, 1990) and to partition net joint moments into their respective muscle forces (Nieminen *et al.*, 1995).

Electromyography can be used in isolation or in combination with other test procedures, and is typically used in conjunction with other measurement systems such as videography, accelerometry, goniometry or isokinetic dynamometry. Although EMG is a useful tool in and of itself, it is frequently necessary to relate the periods of muscle activity temporally to the kinematics of the joint and/or segment of interest. The recording of an electromyographic signal is accomplished with the use of either surface or intramuscular fine-wire electrodes. Surface electromyography, which involves

attaching passive recording electrodes to the skin surface directly above the muscle of interest, is the most common technique because it is non-invasive, easy to administer and the recordings provide an accurate description of the overall level of motor unit recruitment within the muscle. In situations where the muscle of interest lies deep to the skin surface or beneath other muscles, intramuscular electrodes must be used. These electrodes are typically very small, fine-wire electrodes that are inserted into the muscle using a hypodermic needle. Consequently, intramuscular fine-wire EMG is an invasive procedure that detects and provides information about the activation of muscle fibres within a very specific region of the muscle.

Given the structural complexity of the shoulder girdle, most studies that have used EMG to investigate shoulder muscle activity have implemented a combined approach in which both surface and intramuscular fine-wire electrodes were used. Intramuscular electrodes are typically used to record the activity of one or more of the rotator cuff muscles. With this approach researchers have investigated the function of shoulder muscles in athletic skills that are associated with high rates of injury such as swimming and pitching (Jobe *et al.*, 1983; Nuber *et al.*, 1986). With respect to the study of shoulder movement, EMG has been used to measure muscle fatigue (Gerdle *et al.*, 1993; Hägg *et al.*, 1987), to predict muscle force and load distribution patterns among specific muscles in various positions and tasks (Palmerud *et al.*, 1995) and to evaluate the coordination and skill acquisition properties of various upper limb movements (Vorro & Hobart, 1981).

Given the importance of dynamic neuromuscular control to the stability of the shoulder, it is surprising that only a small number of studies have used EMG to

investigate shoulder muscle activity in subjects with glenohumeral instability. Studies of subjects with voluntary posterior instability have shown that abnormal patterns of muscle activation are used to initiate dislocation manoeuvres (Beall *et al.*, 1987; Pande *et al.*, 1989; Rowe *et al.*, 1973). The remaining studies, with one exception (Morris *et al.*, 2004), have investigated muscle activity in subjects with recurrent anterior instability and found a range of movement specific, abnormal levels of activation in several different shoulder muscles (Glousman *et al.*, 1988; Kronberg *et al.*, 1991; McMahon *et al.*, 1996; Wallace *et al.*, 1997). Recently, Morris *et al.* (2004) conducted the first study of shoulder muscle activation in subjects with MDI and found atypical levels of activation in all three sections of the deltoid muscle but found no differences in the activation of the rotator cuff muscles.

Although the general literature on shoulder electromyography is substantial, the specific section pertaining to glenohumeral instability is limited and, until recently, was non-existent for MDI. This is likely because shoulder EMG studies present considerable technical challenges, among them the use of invasive procedures to record the activity of one or more of the rotator cuff muscles. The subscapularis, for example, is particularly difficult to access because it lies in the subscapular fossa between the scapula and the thoracic cage. It stabilizes the glenohumeral joint and contributes to the internal rotation of the humerus but because of its difficult location it has been ignored in almost every electromyographic investigation of the shoulder (Kabada *et al.*, 1992; Németh *et al.*, 1990). Other difficulties associated with using EMG to investigate shoulder instability include the large number of muscles that affect the glenohumeral joint and the difficulty associated with collecting data from all of them simultaneously, as well as the ability of

subjects to move their shoulders through a complete range of motion without pain or apprehension.

Existing studies of shoulder muscle function in glenohumeral instability have quantified the level of muscle recruitment by comparing the amplitude of activation at different stages of the movement cycle (Glousman *et al.*, 1988; Kronberg *et al.*, 1991; McMahon *et al.*, 1996; Morris *et al.*, 2004; Wallace *et al.*, 1997. While recruitment level is an important characteristic of muscle function, an equally important factor is the timing of the activation. Consequently, future studies should investigate both the amplitude and temporal characteristics of shoulder muscle activity in MDI and other types of glenohumeral instability. Future research should also attempt to quantify the kinematics of scapular motion and to examine both the scapulothoracic and scapulohumeral rhythms in an attempt to substantiate the effect produced by these muscles on the skeletal components of the shoulder girdle system.

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3

Repetitive pointing to remembered proprioceptive targets improves 3D hand positioning accuracy

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Published in Human Movement Science 2005; 24: 184-205

Abstract

Repetitive pointing movements to remembered proprioceptive targets were investigated to determine whether dynamic proprioception could be used to modify the initial sensorimotor conditions associated with an active definition of the target position. Twelve blindfolded subjects used proprioception to reproduce a self-selected target position as accurately as possible. Ten repetitions for each limb were completed using overhead and scapular plane pointing tasks. A 3D optical tracking system determined hand trajectory start and endpoint positions for each repetition. These positions quantified three-dimensional pointing errors relative to the target position and the initial and preceding movement repetitions, as well as changes in movement direction and extent. Target position and cumulative start position errors were significantly greater than the corresponding preceding movement (inter-repetition) errors, and increased as the trial progressed. In contrast, hand trajectory start and endpoint inter-repetition errors decreased significantly with repeated task performance, as did movement extent, although it was consistently underestimated for each repetition. Pointing direction remained constant, except for the angle of elevation for scapular plane pointing, which consistently decreased throughout the trial. The results suggest that the initial conditions prescribed by actively defining a proprioceptive target were subsequently modified by dynamic proprioception, such that movement reproduction capability improved with repeated task performance.

3.1 Introduction

The ability to move the arm to position the hand in three-dimensional space represents one of the nervous system's most important achievements. With this ability humans grasp implements and interact with their environment in a limitless number of ways. The processes involved in planning, executing and controlling a multi-link, multiple degree of freedom system with the functional precision necessary to successfully acquire a three-dimensional target are extraordinarily complex. The loss of motor function and level of incapacitation experienced by individuals with Parkinson's disease (Majsak, Kaminski, Gentile, & Flanagan, 1998; O'Suilleabhain, Bullard, & Dewey, 2001; Seidler, Alberts, & Stelmach, 2001) and sensory neuropathy (Gordon, Ghilardi, & Ghez, 1995; Sanes, Mauritz, Dalakas, & Evarts, 1985) attest to the importance of understanding these processes.

The control of coordinated movement is contingent on the delivery of proprioceptive feedback from specialized peripheral receptors (Bevan, Cordo, Carlton, & Carlton, 1994; Cordo, Carlton, Bevan, Carlton, & Kerr, 1994; Gandevia, 1996). Proprioception is required for interjoint coordination and for controlling proximally produced joint interaction torques (Ghez, Gordon, Ghilardi, Christakos, & Cooper, 1990; Sainburg, Poizner, & Ghez, 1993; Sainburg, Ghilardi, Poizner, & Ghez, 1995; Smith & Zernicke, 1987). In a variety of animals such as the cat, proprioceptive afferents regulate rhythmic movements via the influence of central pattern generators (Bässler, 1986; Conway, Hultborn, & Kiehn, 1987; Lennard, 1985; Wolf & Pearson, 1988), and recent evidence suggests this is also the case for walking in humans (Dietz, 2002). With respect

to the control of reaching movements, it has been suggested that motion-dependent proprioception modulates subsequent motor output by updating an internal model of the arm with movement specific limb dynamics information (Ghez *et al.*, 1990; Messier, Adamovich, Berkenblit, Tunik, & Poizner, 2003; Sainburg *et al.*, 1995).

Hand position errors for reaching movements to remembered visually defined (Darling & Miller, 1993; Flanders & Soechting, 1990; Soechting & Flanders, 1989) and proprioceptively defined targets (Adamovich, Berkenblit, Fookson, & Poizner, 1998) indicate that initial limb dynamics estimates may be insufficient and depend on the initial conditions used to define the movement (Hasan, 1992). When pointing to remembered proprioceptive targets (targets that correspond to spatial limb and/or joint positions defined using proprioceptive information) the initial conditions associated with an active definition of the target have been shown to produce greater hand-positioning accuracy than the conditions associated with a passive definition of the target (Adamovich *et al.*, 1998). Actively defined initial conditions presumably contain enough information for the central nervous system (CNS) to successfully accomplish the task. These conditions include a proprioceptive estimate of the spatial limb position and joint configuration, a memorized template of the control signals, and a movement specific estimate of the limb dynamics needed to perform the task (Adamovich *et al.*). However, when performing repetitive pointing movements to an actively defined remembered proprioceptive target, it is not known whether the initial conditions provide the CNS with sufficient information to optimally reproduce the target, or whether motion-dependent proprioception can be used to increase the accuracy of subsequent movements.

Therefore, the purpose of this study was to investigate hand position errors for repetitive pointing movements to remembered proprioceptive targets, to determine whether the initial conditions established by actively defining the target could be improved with motion-dependent proprioception, and if so, to quantify the extent of this improvement. Our expectation was that repeated motion-generated proprioceptive feedback would serve to update and recalibrate the parameters that originally defined the initial conditions of the movement. Consequently, it was hypothesized that hand trajectory endpoint position errors relative to the outcome of the previous movement would be less than the errors measured relative to the target, and that the errors between successive movement repetitions would decrease as further repetitions of the task were performed.

3.2 Materials and Methods

Subjects

Twelve subjects (9 male, 3 female; age = 26.3 ± 5.4 yrs.), with no symptoms or history of neurological disease or motor impairment, participated in the study. The research was conducted following the approval of the authors' institutional ethics review board. Subjects were requested to indicate limb dominance and to provide informed consent prior to participating in the study.

Apparatus

Subjects were seated in the centre of a rectangular test area bounded by four infrared video cameras (one in each corner). The position of the hand in three-dimensional space was quantified using a retroreflective marker (1 cm in diameter) placed on the head of the metacarpophalangeal joint of the middle finger. An optical capture motion analysis system (Vicon Motion Systems, Lake Forest, CA) tracked the three-dimensional movement of the hand with a sampling rate of 60 Hz.

Protocol

The experiment involved performing repetitive, three-dimensional pointing movements to an actively defined remembered proprioceptive (kinesthetic) target. The objective of the experiment was for subjects to return the hand to the target position as accurately as possible without the use of vision (the subjects were blindfolded). The rationale for investigating the reproduction of hand position was based on the results of numerous studies which suggest that the CNS uses a hand-centred coordinate system for the planning and execution of goal-directed reaching movements (Atkeson & Hollerbach, 1985; Brown, Rosenbaum, & Sainburg, 2003b; Flash & Hogan, 1985; Kaminski & Gentile, 1989; Wolpert, Ghahramani, & Jordan, 1995). In addition, the CNS has been shown to selectively minimize hand position error for multijoint arm movements by adjusting the angular configuration of one joint to compensate for variations in the angular configurations of other joints (Adamovich *et al.*, 1998).

Pointing movements to two separate locations of the physical workspace were investigated. The locations were (1) directly overhead and (2) in front of and to the side

of the subject's body (pointing in the scapular plane). Overhead pointing required the arm to be raised to a vertically oriented position above the head and scapular plane pointing consisted of arm elevation in the scapular plane (which is approximately 30° anterior to the coronal plane), with the arm parallel to the horizontal. These tasks were selected because of their familiarity (the subjects did not have to learn any novel tasks or complex motor skills) and generalizability to the repertoire of three-dimensional reaching movements routinely performed in everyday life (reaching to grasp objects above the body and at various positions in front of and lateral to the body). For each randomly assigned task, two trials were performed, one with the left arm and one with the right (limb order was also randomized). For each task trial, the subjects were instructed to return the hand to the target position 10 times in succession. In total, each subject performed 40 hand-repositioning movements (2 tasks x 2 limbs x 10 repetitions per trial).

The rationale for pointing to a remembered proprioceptive target (as opposed to a remembered visual target) was to investigate the effect of dynamic proprioceptive feedback on the motor output of successive movements. Selecting a proprioceptive target ensured that the CNS possessed only a proprioceptive (and not a visual) representation of the joint configuration and spatial orientation of the limb. The process of actively defining a proprioceptive target was used so that proprioceptive information concerning the static forces and joint torques needed to counter the effect of gravity while maintaining the limb at the target position would be available to the CNS. Actively selecting the target also ensured that the CNS possessed an initial motion-dependent proprioceptive estimate of the limb dynamics needed to perform the task. To this end, active "virtual" proprioceptive targets were defined. The target positions were "virtual"

because no actual physical target was used. Instead, subjects actively moved the arm to either of the two physical workspace locations on the ipsilateral side of the body and then self-selected a position in space with the hand. The subject and the optical tracking system defined the coordinates of this hand position as the "target".

Methods

Movement demonstrations and verbal instructions concerning the specifics of the test procedure were given to each subject at the beginning of the test session. Subjects were instructed to move the upper limb in a specified manner and to perform a minimum of three practice repetitions until they were comfortable with each task. Practice repetitions were performed at the beginning of each trial after the subjects were blindfolded. Immediately following the practice repetitions, the subjects were instructed to move the limb to the appropriate workspace location (overhead or scapular plane) and to select their virtual target position. The limb was held against gravity with the hand in the target position for several seconds. Following this delay, the subjects were instructed to begin the trial. The arm was lowered to an area located along the midline of the ipsilateral thigh just below the hip and medial to the greater trochanter before being returned to the target position. This position, which subjects were not instructed to accurately reproduce, served as the start position for each movement trajectory. No performance feedback of any kind was given concerning the outcome or accuracy of each movement repetition or trial. Subjects were instructed not to move the wrist, to avoid full elbow extension and to self-select a comfortable movement speed that would enable them to reproduce the target position with the greatest accuracy. Subjects were also instructed not to count, to

complete all ten repetitions of the trial continuously and were told by the experimenter when to stop.

Data Analysis

Three-dimensional hand marker position data were smoothed with a zero-lag, 2nd order Butterworth low-pass digital filter at a cutoff frequency of 6 Hz (Winter, 1990) using MATLAB (The Mathworks Inc., Natick, MA). This procedure smoothed the trajectory so that any subjectivity in selecting the appropriate video frames, which corresponded to the start and endpoint positions of each repetition, was eliminated. Start and endpoint positions were determined by graphically viewing the three-dimensional hand marker trajectory to obtain a visual estimate of the direction reversal points. To corroborate the values determined by the trajectory reversal method, start and endpoint positions were also obtained using the minima of the three-dimensional tangential hand velocity profile.

To assess the effects of dynamic proprioception on hand-positioning accuracy, two aggregate three-dimensional (x, y and z directions combined) trajectory endpoint position error measures were calculated. The target position error (TPE) quantified the difference between each trajectory endpoint position and the proprioceptive target, which provided a relative measure of hand-positioning accuracy with respect to the initial set of movement conditions (obtained by actively defining the target). The inter-repetition error (IRE) quantified the difference between each endpoint position and the previous endpoint position, which provided a relative measure of hand-positioning accuracy with respect to the most recent or updated set of movement conditions. Recent studies of repetitive proprioceptive arm movements, similar to those presented here, have shown that the

position errors of planar pointing trajectories accumulate (or drift) over time with respect to the location of the first trajectory (Brown, Rosenbaum, & Sainburg, 2003a; Brown *et al.*, 2003b). To assess whether the three-dimensional hand trajectory position errors accumulated over time, the cumulative (CSE) and instantaneous (ISE) start position errors were also determined. The cumulative start position error quantified the difference between each trajectory start position and the initial start position, whereas the instantaneous error assessed the difference between each start position and the preceding start position. The formulae for calculating the mean TPE, IRE, CSE and ISE for each trial were:

$$\text{Mean TPE} = \left(\sum_{n=1}^{10} \sqrt{(X_n - X_o)^2 + (Y_n - Y_o)^2 + (Z_n - Z_o)^2} \right) \div 10$$

$$\text{Mean IRE} = \left(\sum_{n=1}^{10} \sqrt{(X_n - X_{n-1})^2 + (Y_n - Y_{n-1})^2 + (Z_n - Z_{n-1})^2} \right) \div 10$$

$$\text{Mean CSE} = \left(\sum_{n=2}^{9} \sqrt{(X_n - X_1)^2 + (Y_n - Y_1)^2 + (Z_n - Z_1)^2} \right) \div 9$$

$$\text{Mean ISE} = \left(\sum_{n=2}^{9} \sqrt{(X_n - X_{n-1})^2 + (Y_n - Y_{n-1})^2 + (Z_n - Z_{n-1})^2} \right) \div 9$$

where X, Y and Z represent global coordinate positions in 3D orthogonal space, o = the target position coordinate and n = the repetition number. These aggregate error calculation methods were used so that the error term for each direction would contribute equally to the total, so that the location of the point in space with respect to the origin of the axis did not bias the result.

Hand trajectory direction and extent were also determined to investigate whether there was any change in these parameters throughout the course of the trial. Recent

evidence has suggested that the CNS preserves the direction and extent of planar reaching movements in the absence of vision by using motion-dependent proprioception to modulate joint muscle torque (Brown *et al.*, 2003a). Movement extent was determined by calculating the three-dimensional radial distance vector of each trajectory (see Fig. 3.1). The formula for calculating extent was:

$$\text{Extent } (E) = \sqrt{(Xe - Xs)^2 + (Ye - Ys)^2 + (Ze - Zs)^2}$$

where e = the endpoint position coordinate and s = the start position coordinate. The percent change in movement extent for each repetition was calculated by determining the difference in magnitude between each three-dimensional radial distance vector and the first (cumulative) or preceding (instantaneous) vectors:

$$\% \text{ change in cumulative } E = \left\{\frac{(E_n - E_1)}{E_1}\right\} * 100$$

$$\% \text{ change in instantaneous } E = \left\{\frac{(E_n - E_{n-1})}{E_{n-1}}\right\} * 100$$

where E = extent and n = the repetition number.

The direction of each pointing trajectory was determined by measuring the angle of rotation of the three-dimensional radial distance vector about the pitch (Y) and yaw (Z) axes (see Fig. 3.1). The angle about the roll axis (the X-axis, which corresponded to the longitudinal axis of the arm) was not calculated because it did not provide any additional

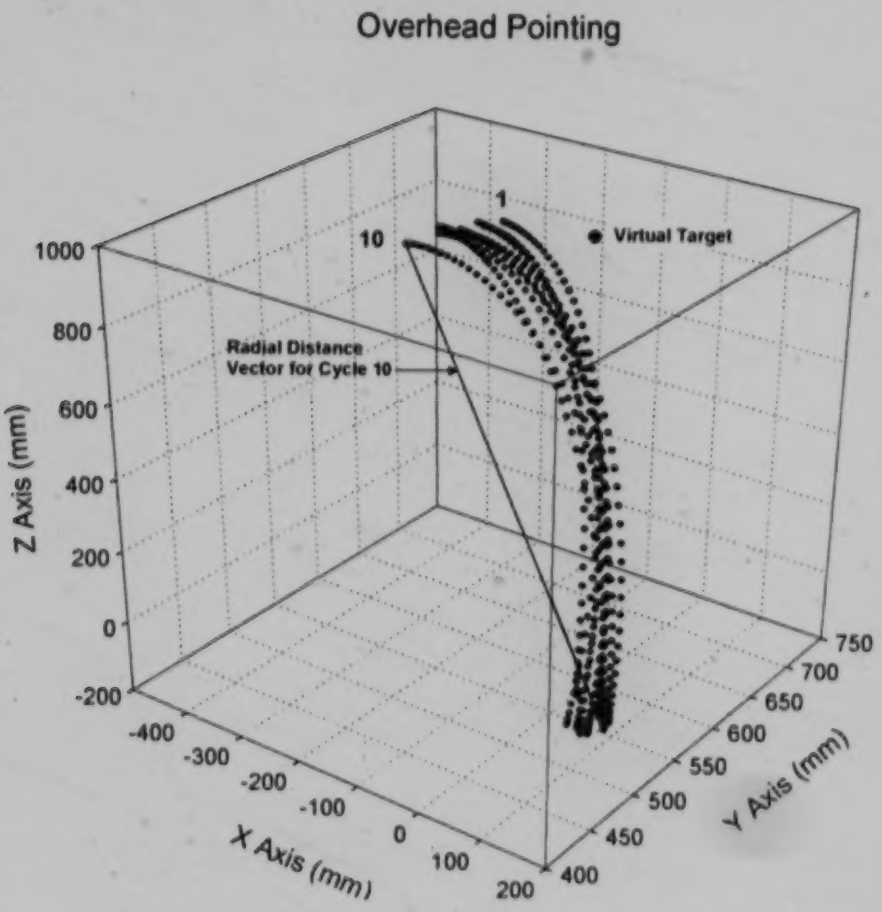

Figure 3.1: *Representative three-dimensional trajectories for an overhead pointing trial. Note the extent to which the endpoint position of the first repetition (1) missed the target and the proximity of subsequent trajectories to the initial movement, rather than to the proprioceptive target. The plot also demonstrates the tendency of the endpoint positions to drift from the target (1 = repetition = 1, 10 = repetition = 10).*

information with respect to limb movement direction. The pitch angle defined the angle of elevation of the radial distance vector with respect to the horizontal plane (*XY*), while the yaw angle defined the angle of horizontal abduction/adduction with respect to the sagittal plane (*XZ*). As with movement extent, the cumulative and instantaneous changes in pitch and yaw angles for each repetition were determined throughout the trial.

Two-way (task x dominance) analyses of variance (ANOVA) with repeated measures (repetition) were used to examine the effect of the repetition number on the dependent variables for endpoint position error (TPE and IRE), start position error (CSE and ISE), movement direction (pitch and yaw angles) and extent. Because limb dominance was not controlled for experimentally, it was included in the statistical model to account for any potential effects this might have on the results. Mean TPE (averaged across all trials and repetitions) vs. mean IRE, and mean CSE vs. mean ISE were also compared using a paired Student's T-test. *P* values less than .05 were reported as statistically significant. Pair-wise comparisons between each repetition error and the first repetition error were performed using Bonferroni adjusted *p* values of .0056 (.05/9) and .00625 (.05/8), for TPE and IRE (9 comparisons) and CSE and ISE (8 comparisons), respectively.

3.3 Results

Endpoint Position Errors

The data show that target position and hand inter-repetition errors varied as subjects repeatedly pointed to a remembered proprioceptive target. The repeated measures

ANOVA revealed that the repetitive nature of the task had a significant effect on both the target position error (TPE), F (9, 43) = 9.06, $p < .0001$, and the inter-repetition error (IRE), F (9, 43) = 12.18, $p < .0001$. There were no significant within-group interaction effects for either TPE or IRE. Endpoint position error relative to the target increased for both tasks as the trial progressed (Fig. 3.2). In contrast, endpoint inter-repetition error decreased and was then maintained throughout the trial for both tasks following the third repetition (Fig. 3.2). Task type (overhead or scapular plane) had no effect on either the target position error, F (1, 43) = 0.31, $p < .58$, or the inter-repetition error, F (1, 43) = 3.04, $p < .09$. Similarly, limb type (dominant or non-dominant) had no effect on TPE, F (1, 43) = 0.01, $p < .94$, or IRE, F (1, 43) = 0.11, $p < .74$. In addition, there were no significant interaction effects (task x dominance) for either TPE, F (1, 43) = 0.27, $p < .61$, or IRE, F (1, 43) = 0.00, $p < .99$.

Post-hoc comparisons for TPE (9) indicated that repetitions 4, 5, 8, 9 and 10 were significantly greater than repetition 1 ($p < .0056$). For IRE, the second inter-repetition error was significantly less than the first IRE and the third inter-repetition error was significantly less than the second ($p < .05$). All subsequent inter-repetition errors were significantly less ($p < .0056$) than the first IRE (the error between the target and the first repetition). The data indicate that for both tasks, endpoint position errors relative to the target accumulated as the trial progressed, whereas endpoint position with respect to the previous repetition improved to a new level over the first three repetitions, which was then maintained throughout the rest of the trial.

In comparing mean TPE to IRE, it was found that mean TPE was significantly greater ($p < .0001$) than mean IRE (see Fig. 3.4). In 44 out of the 47 trials (12 subjects x 2

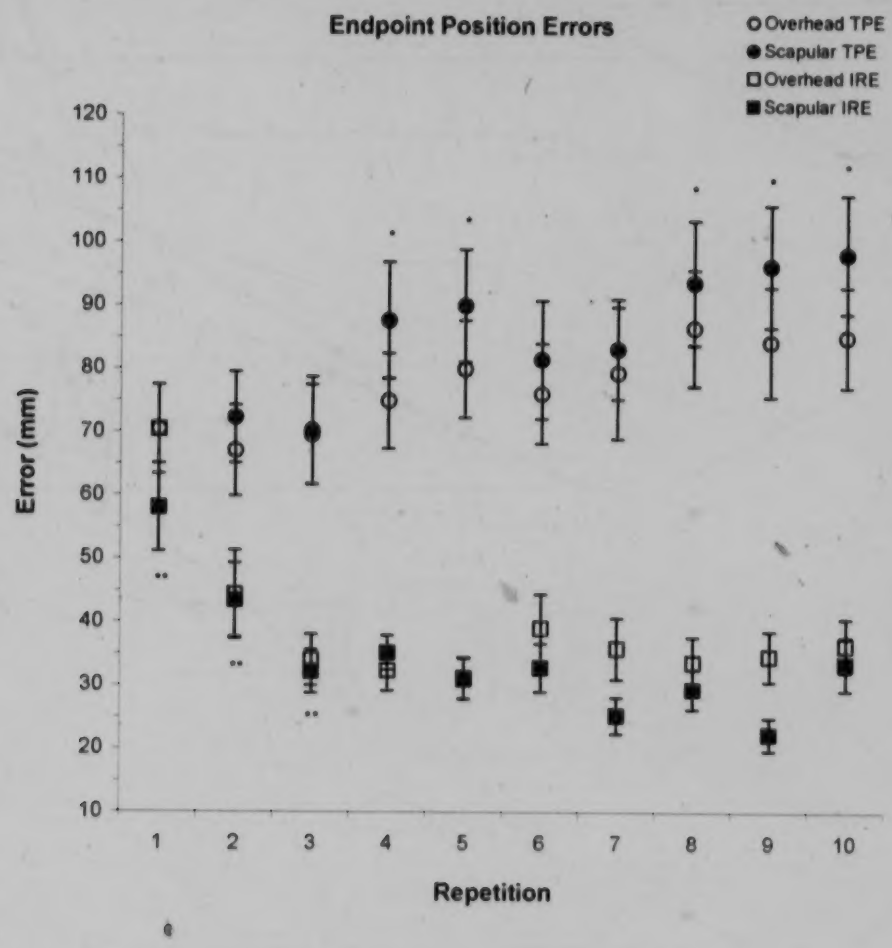

Figure 3.2: *Endpoint position errors for overhead and scapular plane pointing. TPE = target position error, IRE = inter-repetition error. * = TPEs significantly different ($p < 0.0056$) from the first TPE (repetition 1) and ** = IREs significantly different ($p < 0.05$) from the previous IRE. All IREs were significantly different from the first IRE ($p < 0.0056$). Error bars indicate ± 1 standard error of the mean.*

limbs x 2 movements = 48 trials; one trial was discarded because of a marker reconstruction problem), mean TPE was greater than mean IRE (93.6%).

Start Position Errors

As with the endpoint position errors, the cumulative (CSE) and instantaneous start position errors (ISE) changed significantly as the tasks were repeated. Significant within-group main effects were present for both CSE, F (8, 43) = 6.28, $p < .0001$, and ISE, F (8, 43) = 2.19, $p < .05$. None of the within-group interaction effects for CSE or ISE were significant. The cumulative and instantaneous start position errors increased and decreased, respectively, over the course of the trial (see Fig. 3.3). The corroboration of the pattern of errors to the endpoint position errors (compare Figures 3.2 and 3.3) demonstrates that the absolute spatial position of the pointing trajectories drifted over time (see also Fig. 3.1), and that the degree of inter-trajectory drift at the end of the trial was less than at the beginning. Cumulative start position errors were not affected by either task type (overhead or scapular plane), F (1, 43) = 0.37, $p < .54$, or limb type (dominant or non-dominant), F (1, 43) = 0.68, $p < .42$. Similarly, instantaneous start position errors for overhead and scapular plane pointing were not significantly different, F (1, 43) = 0.13, $p < .72$, however the errors between dominant and non-dominant limbs were, F (1, 43) = 4.92, $p < .03$, such that mean ISE was less for the dominant limb. No significant interaction effects (task x dominance) were found for either CSE, F (1, 43) = 0.02, $p < .89$, or ISE, F (1, 43) = 0.25, $p < .62$.

For the cumulative start position error, post-hoc comparisons (8) revealed that all CSEs, with the exception of the second CSE (which approached significance at

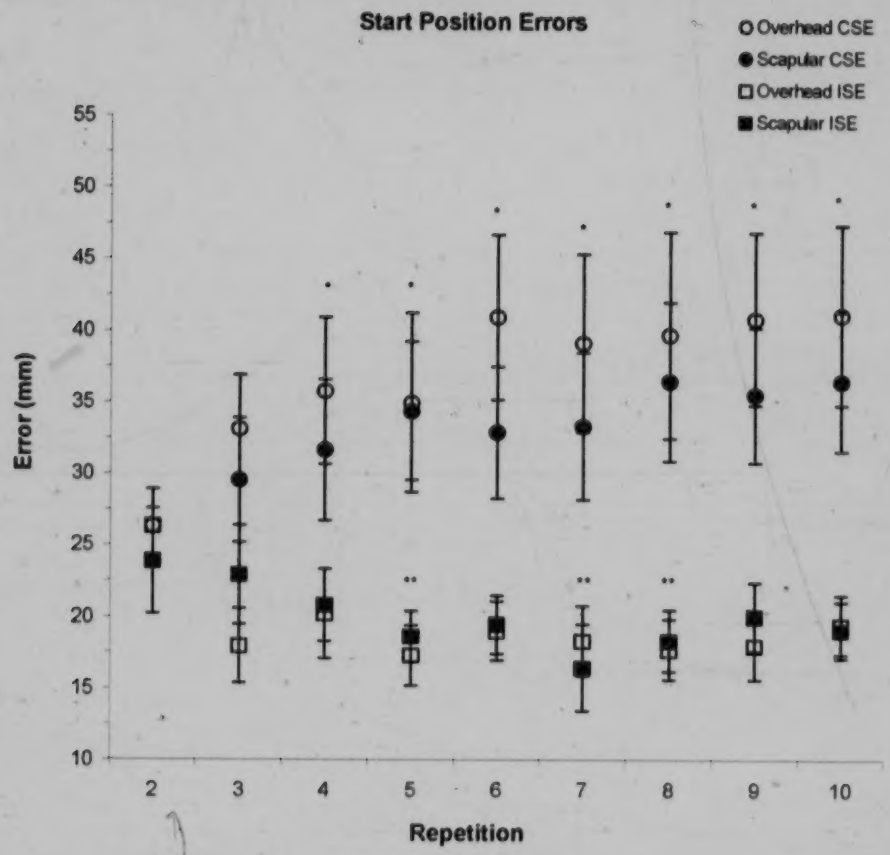

Figure 3.3: *Start position errors for overhead and scapular plane pointing. CSE = cumulative start position error, ISE = instantaneous start position error. * = CSE's significantly different ($p < 0.00625$) from the first CSE, and ** = ISE's significantly different from the first ISE ($p < 0.00625$). None of the ISEs were significantly different from any of the previous ISEs. Error bars indicate ± 1 standard error. Note the similarity of the pattern of errors to those shown in Fig. 3.2 for the trajectory endpoints.*

$p < .0067$), were significantly greater ($p < .00625$) than the first CSE (Fig. 3.3). For ISE, no significant differences were found between any of the ISEs and the previous ISE,

however the differences between ISEs 2 and 5, 2 and 7 and 2 and 8 were significant ($p <$ .00625). The results indicate that for both tasks, start position errors relative to the first trajectory increased as the trial progressed, and that over the course of several repetitions, the instantaneous start position error improved relative to the beginning of the trial. As with the endpoint position error, it was found that the mean cumulative start position error was significantly greater ($p < .001$) than the mean instantaneous error for both tasks (see Fig. 3.4).

Movement Direction and Extent

As with the start and endpoint position errors, both the cumulative (relative to the first trajectory) and instantaneous (relative to the preceding trajectory) changes in movement direction and extent were investigated for each repetition (cycle). The cumulative changes in extent (CCE) shown in Fig. 3.5 (Panel A) indicate that the radial distances of each trajectory were less than the first trajectory (negative % change), and that these reductions were consistent and gradually accumulated over the course of the trial. The repeated measures ANOVA demonstrated that these accumulated changes in extent were significant, $F(8, 43) = 5.13$, $p < .0001$. To further substantiate the accumulated deficit in movement extent, the CCE for repetition 10 was compared to the CCE for repetition 1 and was found to be significant ($p < .02$), indicating that movement extent, relative to the first repetition, was underestimated more at the end of the trial than at the beginning. No significant within-group interaction effects or between-group effects were found for CCE (task: $F(1, 43) = 1.50$, $p < .23$; limb: $F(1, 43) = 0.24$, $p < .63$; task x limb: $F(1, 43) = 0.03$, $p < .86$), indicating that task and limb type had no effect on the accumulated

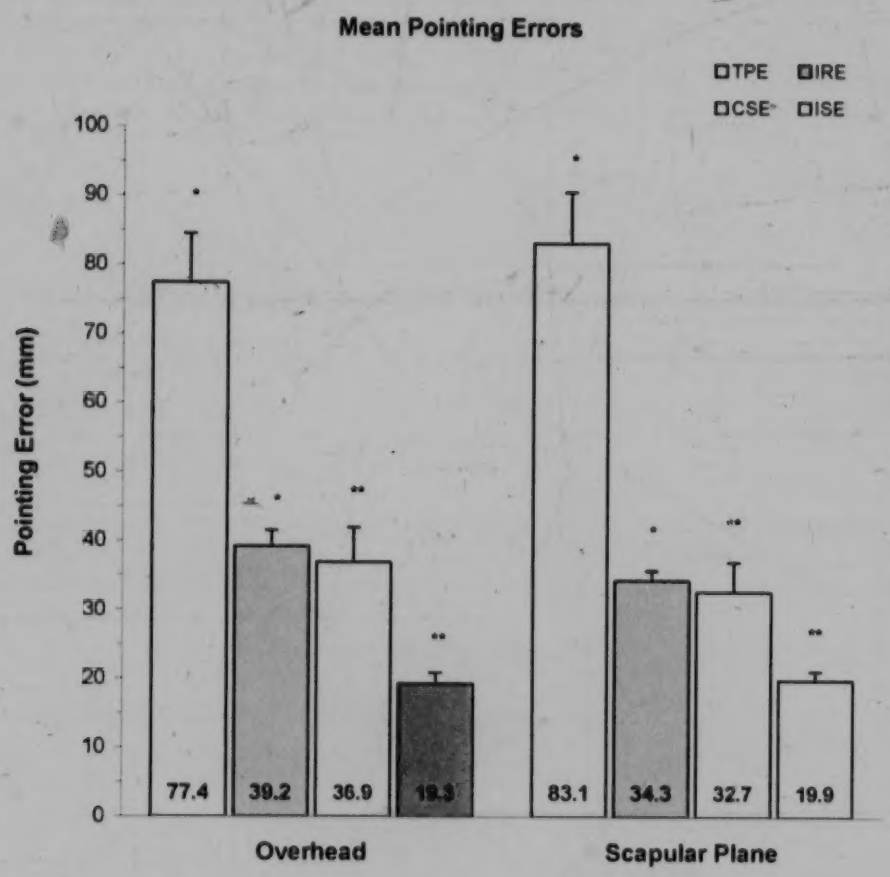

Figure 3.4: *Mean pointing errors (endpoint and start position errors) for all scapular plane and overhead trials. * = significant difference ($p < 0.0001$) for TPE vs. IRE and ** = significant difference ($p < 0.001$) for CSE vs. ISE. Error bars indicate ± 1 standard error.*

changes in movement extent.

The instantaneous change in extent (ICE) results demonstrated that there were no

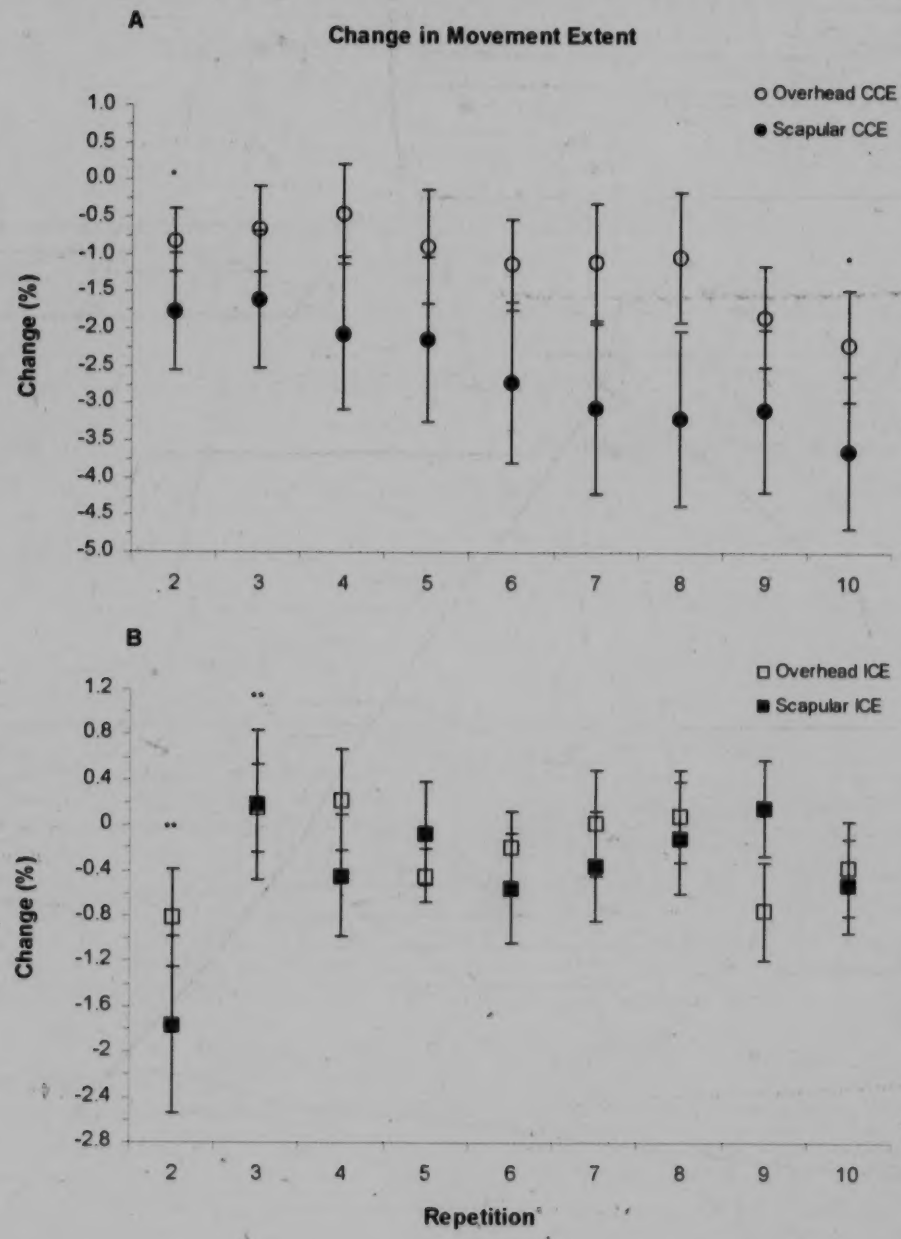

Figure 3.5: *Cumulative (A) and instantaneous (B) changes in movement extent for overhead and scapular plane pointing. CCE = cumulative change in extent, ICE = instantaneous change in extent. * = significant difference (p < 0.003) for CCE 2 vs. CCE*

*10. ** = significant difference ($p < 0.02$) for ICE 2 vs. ICE 3. Error bars indicate ± 1 standard error.*

significant within (cycle: F (8, 43) = 1.37, $p < .21$; cycle x limb: F (8, 43) = 1.05, $p < .40$; cycle x task: F (8, 43) = 0.64, $p < .75$; cycle x limb x task: F (8, 43) = 0.77, $p < .63$) or between-group effects (limb: F (1, 43) = 0.08, $p < .78$; task: F (1, 43) = 1.05, $p < .31$; limb x task: F (1, 43) = 0.03, $p < .87$) for this variable. However, the data in Fig. 3.5 (Panel B) indicate that the first ICE (ICE 2) was much larger than the second ICE (ICE 3) and the remaining ICEs of the trial. A post-hoc comparison between ICEs 3 and 2 revealed that the difference was significant ($p < .02$), which demonstrated that the change in movement extent between repetitions 3 and 2 was significantly less than the change between repetitions 2 and 1 (Fig. 3.5B). The data indicate that in terms of extent, trajectory 3 was a better reproduction of the preceding trajectory than trajectory 2.

With respect to the direction of pointing, there was a significant within-group main effect for the cumulative change in pitch angle (CPA), F (8, 43) = 4.29, $p < .0001$, as well as a significant cycle x task interaction, F (8, 43) = 3.43, $p < .001$. The mean cumulative changes in pitch angle for each repetition are presented in Fig. 3.6 (Panel A). The data show that the angle of elevation about the pitch axis for overhead pointing was maintained throughout the trial, whereas for scapular plane pointing it was not. For the scapular plane task, a post-hoc comparison of CPA 10 vs. CPA 2 indicated that there was a significant drop in trajectory elevation as the task was repeated ($p < .002$). There were no other significant within-group effects for CPA, however, there was a significant between-group main effect for task, F (1, 43) = 3.94, $p < .05$, indicating that the change

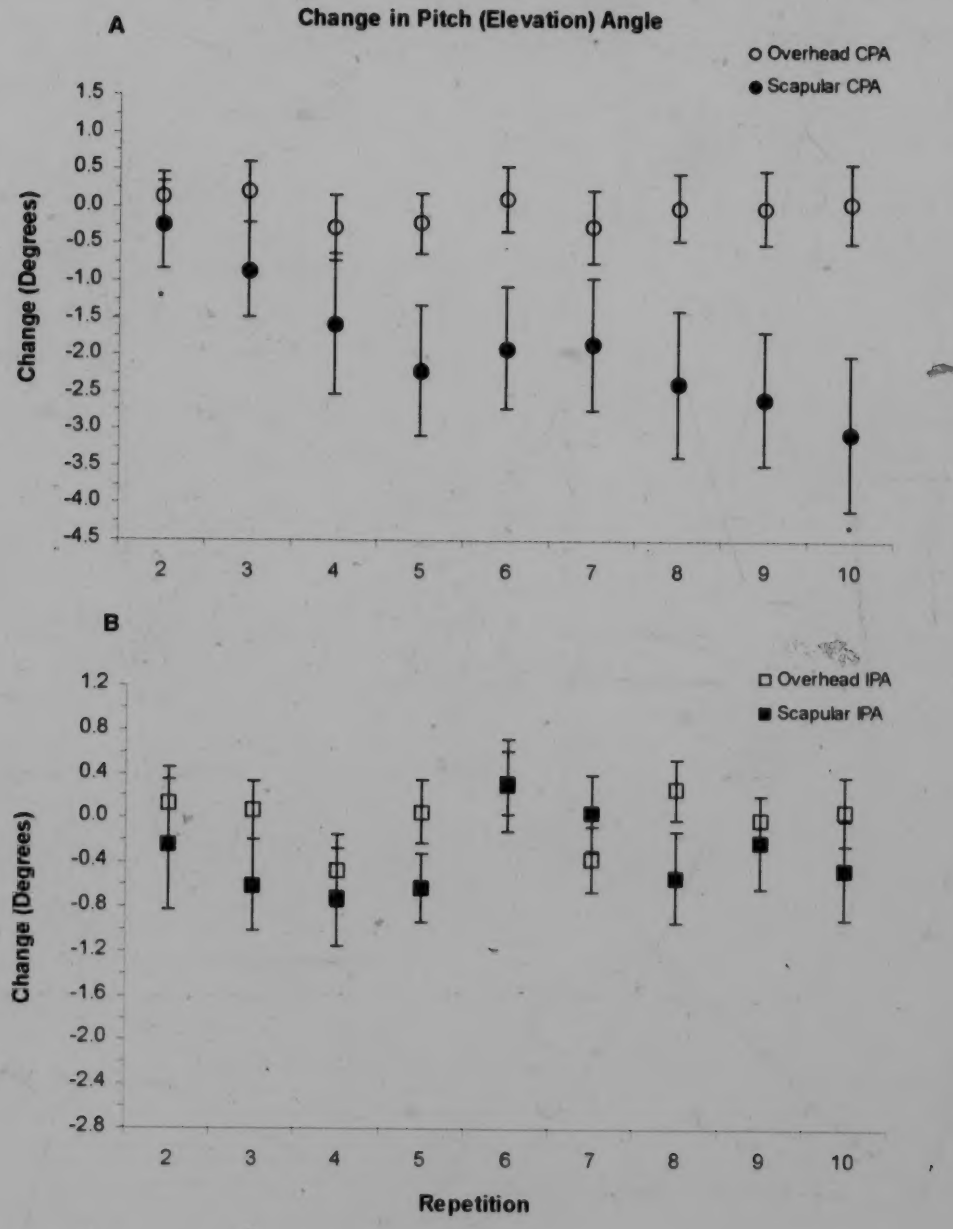

Figure 3.6: *Cumulative (A) and instantaneous (B) changes in pitch (elevation) angle for overhead and scapular plane tasks. CPA = cumulative change in pitch angle, IPA =*

*instantaneous change in pitch angle. * = significant difference ($p < 0.002$) for CPA 2 vs. CPA 10 for scapular plane pointing. Error bars indicate ± 1 standard error.*

in angle for scapular plane pointing was significantly greater than the change in angle for overhead pointing. There were no other significant between-group effects for CPA (limb: $F(1, 43) = 0.48, p < .49$; limb x task: $F(1, 43) = 0.05, p < .82$).

The data for the instantaneous change in pitch angle (IPA) are presented in Fig. 3.6B. The statistical analysis revealed that there were no significant within-group effects for IPA (cycle: $F(8, 43) = 0.78, p < .62$; cycle x limb: $F(8, 43) = 0.58, p < .79$; cycle x task: $F(8, 43) = 0.52, p < .85$; cycle x limb x task: $F(8, 43) = 0.66, p < .73$). For the scapular plane task, almost all the instantaneous changes in pitch angle were negative (Fig. 3.6B), which produced the accumulated decline in elevation shown in Panel A (Fig. 3.6). As with the cumulative results, there was a significant between-group main effect for task type, $F(1, 43) = 6.49, p < .01$, demonstrating that there was a difference in the change in instantaneous elevation angle between the two tasks (the change in the instantaneous angle of elevation was greater for scapular plane pointing). No other significant between-group effects were found for IPA (limb: $F(1, 43) = 0.11, p < .74$; limb x task: $F(1, 43) = 0.04, p < .84$).

For the cumulative (CYA) and instantaneous (IYA) changes in angle about the yaw axis (Fig. 3.7A and B, respectively), no significant within-group differences were found for either CYA (cycle: $F(8, 43) = 0.88, p < .54$; cycle x limb: $F(8, 43) = 0.72, p < .67$; cycle x task: $F(8, 43) = 0.94, p < .48$; cycle x limb x task: $F(8, 43) = 0.68, p < .71$) or IYA (cycle: $F(8, 43) = 1.02, p < .42$; cycle x limb: $F(8, 43) = 0.53, p < .83$; cycle x task:

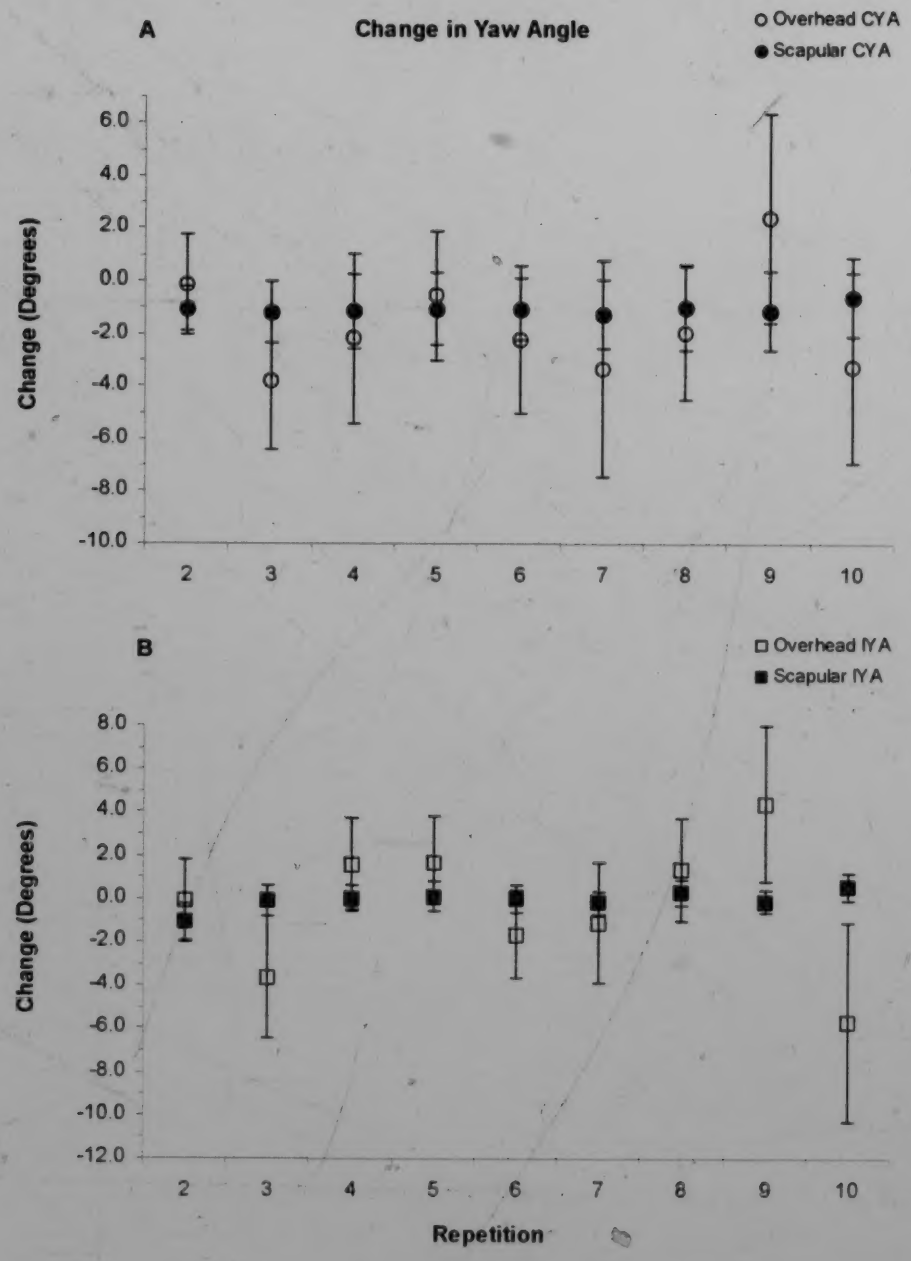

Figure 3.7: *Cumulative (A) and instantaneous (B) changes in yaw angle for overhead and scapular plane tasks. CYA = cumulative change in yaw angle, IYA = instantaneous*

change in yaw angle. There were no significant cumulative or instantaneous changes in yaw angle for either task. Error bars indicate ± 1 standard error.

F (8, 43) = 1.18, $p < .31$; cycle x limb x task: F (8, 43) = 0.58, $p < .79$). Similarly, there were no significant between-group effects for either CYA (limb: F (1, 43) = 0.08, $p < .78$; task: F (1, 43) = 0.06, $p < .80$; limb x task: F (1, 43) = 0.65, $p < .42$) or IYA (limb: F (1, 43) = 0.70, $p < .41$; task: F (1, 43) = 0.54, $p < .47$; limb x task: F (1, 43) = 1.11, $p < .30$). Consequently, the results demonstrate that pointing direction about the yaw axis (angle of horizontal abduction/adduction of the arm) was consistently maintained throughout the course of the trial, particularly for scapular plane pointing, which was performed with very low inter-trial variability (Fig. 3.7A and B).

3.4 Discussion

The purpose of this study was to investigate hand trajectory endpoint positioning for three-dimensional repetitive pointing tasks to remembered proprioceptive targets, to determine whether motion-dependent proprioception could improve positioning accuracy relative to the initial conditions of the movement. The initial conditions consisted of a static proprioceptive estimate of the target hand position, of the associated limb configuration, and of the control signals and movement specific limb dynamics of the task, and were prescribed through the process of actively defining the target. Subjects pointed to remembered proprioceptive targets so that vision would not have any effect on movement outcome. The results suggest that repeated motion-generated proprioceptive

feedback was used to update and recalibrate the parameters that originally defined the initial conditions of the movement. The hypothesis that the target position errors, which evaluated movement outcome in terms of the task objective and the initial conditions present at the beginning of the trial, would be greater than the inter-repetition errors, which evaluated movement outcome in terms of the most recent set of movement instructions and conditions, was supported. It was also hypothesized that the hand trajectory endpoint inter-repetition errors would decrease as further repetitions of the task were completed. The endpoint inter-repetition error results demonstrated that the accuracy of endpoint positioning relative to the outcome of the preceding movement improved in two stages, after the first repetition and again after the second repetition. The results also showed that after the third repetition, this improved level of endpoint positioning was maintained throughout the rest of the trial. Because the primary objective of this study was to investigate the outcome of movements directed towards a remembered proprioceptive target, the hand trajectory endpoint position errors will be discussed first, followed by a discussion of the trajectory start position errors, and finally by a discussion of the observed changes in movement direction and extent.

Endpoint Position Errors

The results demonstrated that the mean and individual target position errors were greater than the mean and individual inter-repetition errors for both overhead and scapular plane pointing. For each repetition of the ten-cycle trial, the hand was positioned closer to the location of the previous movement than to the intended target. As the trial progressed the target position error increased while the inter-repetition error decreased considerably,

such that the inter-repetition error between positions two and one was less than the error between position one and the target. Similarly, the error between positions three and two was less than the error between positions two and one. The decrease in the second and third inter-repetition errors suggests that control signals for these repetitions continued to evolve in response to updated proprioceptive information acquired during the course of the previous movement. Beyond this point (repetition three), hand inter-repetition error did not change, suggesting that an optimal or terminal level of performance had been reached. The results of this study suggest that at least two iterations of the sensorimotor feedback control loop are necessary before maximum movement reproduction capability can be achieved in the absence of vision.

The results demonstrate that the original sensory and motor specifications of the target, which defined the initial conditions prior to the execution of the first repetition, became increasingly less relevant as the movement was repeated, suggesting that these specifications either faded or were updated several times before an optimal level of execution was achieved. The current findings are consistent with the idea that feedforward mechanisms use the presence and availability of motion-dependent proprioceptive input to modify motor output (Ghez *et al.*, 1990; Messier *et al.*, 2003; Sainburg *et al.*, 1995). According to this theory, the reduction in inter-repetition error for the second and third repetitions occurred because dynamic proprioceptive input provided updated endpoint position and/or limb dynamics information to a central internal model of the task. It is important to note that these improvements occurred after the subjects had performed a minimum of three practice repetitions, which concluded (approximately 15 seconds) immediately prior to the beginning of each trial. The only difference between

the practice repetitions and the test repetitions was that subjects did not self-select a virtual target position before they practiced the task. Practice repetitions were included as part of the experimental protocol to ensure that all subjects were comfortable with the tasks and were able to perform them correctly. It was not possible to establish whether the practice repetitions had any effect on the position error results, in particular the target position error for the first repetition, because the trajectories and endpoint positions of these movements were not recorded.

Recent studies into the distributed nature of the motor control system suggest that the cerebellum receives both sensory and motor inputs to create a forward model that makes predictions about the sensory outcome of movements (Miall & Reckess, 2002). In the Smith Predictor model of motor control (Miall, Weir, Wolpert, & Stein, 1993), the cerebellum receives a copy of the motor commands (an efference copy) along with the reafferent sensory input generated by the movement. The cerebellum uses this information to produce a forward model whose output provides an estimate or prediction of the future sensory state of the system. The accuracy of this state estimate is then evaluated by comparing it to the post-execution state (the actual outcome) when new sensory reafferent information becomes available (Miall & Reckess, 2002). Although this study does not address the question of whether the cerebellum is the structure responsible for creating an internal forward model, the results do support the presence of such a model and suggest that motion-generated proprioception (sensory reafference) adapts motor output by providing the model with updated movement and/or position related information. The inter-repetition error results, which demonstrated recurred improvement followed by a constant level of accuracy, suggest that this updating process is iterative in

nature, such that control signals are incrementally fine-tuned to the point where the system achieves an optimal or steady state.

Equally relevant is the likelihood that the specific conditions associated with pointing to a remembered proprioceptive target influenced the results. Studies have shown that the CNS depends on visual input for feedforward control and the planning of movement trajectories (Ghez *et al.*, 1990) and dominates over proprioception when performing target-pointing movements (Taylor & McCloskey, 1988). Viewing static hand position prior to initiating non-visually guided movements improves pointing accuracy compared to non-viewing conditions, presumably by permitting a better encoding of the initial conditions of the motor apparatus (Desmurget, Rosetti, Jordan, Meckler, & Prablanc, 1997). These and other results (Rosetti, Stelmach, Desmurget, Prablanc, & Jeannerod, 1994) suggest that visual input is critical for optimizing feedforward mechanisms via the pre-movement encoding of limb and target positions. Because vision was excluded in this study, it is possible that the CNS did not possess sufficient information about the target position or the initial limb configuration, and consequently was unable to obtain a satisfactory estimate of the initial conditions of the task. There is also evidence to suggest that, without vision, static proprioception alone is not able to provide the CNS with a sufficient estimate of the initial limb configuration for the purpose of attaining a spatial target (Desmurget *et al.*, 1997). Consequently, an insufficient definition of the target position and inadequate estimate of the initial conditions of the movement may have contributed to the large magnitude of the target position error for the first repetition.

Start Position Errors

The findings show that the pattern of errors (cumulative and instantaneous) for the hand trajectory start positions mirrored the pattern of errors (target and inter-repetition) for the trajectory endpoint positions. Although different terminology was used to refer to errors at opposite ends of the trajectory (for the sake of clarity), the conceptual basis for the errors and the method of calculating them was the same. The fundamental difference between the start position and endpoint position errors was that hand position reproduction at the beginning of each trajectory was not an objective of the task, and therefore did not involve a pre-trial, actively defined proprioceptive target. Nevertheless, over the course of five repetitions the instantaneous start position error decreased significantly, suggesting that dynamic proprioceptive input was used to improve the reproduction of the entire trajectory, and not just the endpoint position.

The target position and cumulative start position errors demonstrated that the three-dimensional trajectories of both tasks drifted with respect to their initial positions over the course of the trial. These findings are consistent with other studies that have reported error congruency between trajectory start and endpoint positions (Bock & Arnold, 1993; Bock & Eckmiller, 1986). The current results also provide support for the study of Brown *et al.* (2003a), who found that the start location for repetitive planar reaching movements performed without vision drifted on average 8 cm from the original position. The magnitude of start position drift in the current study (3.47 cm) was considerably less than that reported by Brown *et al.*, and most likely occurred because there were fewer repetitions for error to accumulate (10 versus 75). The data also showed that the amount of endpoint position drift for both tasks was greater than the degree of start position drift

(2.3 times larger; 80.3 mm vs. 34.7 mm). The most likely explanation for this difference is that the degree to which the target was missed on the first repetition (mean of 64 mm) biased the comparison. Given that all pointing errors were calculated using a three-dimensional aggregate formula, the concept of drift (the accumulation of pointing error) in this context indicates a collective or net increase in pointing error across all three directions.

Movement Direction and Extent

The results for pointing direction indicated that, with one exception, hand trajectory direction was maintained throughout the trial. In the case of the angle of elevation for scapular plane pointing, which was not maintained, the results show that the decline in elevation occurred gradually and did not involve any instantaneous changes between repetitions. Notwithstanding this exception, the current results are consistent with the recent work of Brown *et al.* (2003a, 2003b), who found that the direction of pointing for repetitive proprioceptive arm movements to remembered visually defined targets was preserved despite hand trajectory drift. However, because the subjects in these studies performed two-dimensional arm movements in the horizontal plane, there was no direction of pointing in the sagittal plane to assess. Given that the current study assessed three-dimensional arm movements, it is conceivable that gravity could have been responsible for the gradual and persistent decline in the elevation of scapular plane pointing. Because the endpoint position of this task was associated with a shoulder flexion angle of approximately 90 degrees, it is likely that the gravitational torque at the shoulder was greater than it was for the overhead task, in which the arm was oriented

parallel to, rather than perpendicular to, the force of gravity. Consequently, the cumulative decline in elevation angle could have occurred because the CNS consistently underestimated the amount of torque needed to counteract the effect of gravity in the endpoint position.

With respect to movement extent, the results also showed that repetitive pointing was associated with cumulative and instantaneous changes in movement extent. The cumulative changes in movement extent indicate that trajectory distance was consistently underestimated throughout the trial, and that this systematic underestimation produced a significantly abbreviated trajectory by the end of the trial. It is also remarkable that although extent was consistently underestimated, the degree to which it was underestimated was not. Specifically, the instantaneous results demonstrated that the second movement reproduction (i.e., repetition #3) was a much better estimate of extent than the first (repetition #2). It is interesting to note that the findings and general pattern of the extent data were conspicuously similar to the start and endpoint position error data. This pattern consisted of a relatively poor estimate for the first movement reproduction (i.e., repetition #2), followed immediately by a significantly improved estimate that was then maintained, with minor imperfections, throughout the rest of the trial. The cumulative trajectory position and extent data indicate that the consistency of these minor imperfections was responsible for the accumulated changes that occurred over the course of the trial. It is also interesting to note that the instantaneous improvement in the estimation of extent occurred despite the fact that subjects were not specifically instructed to accurately reproduce the trajectory start position.

To further investigate changes in movement extent, it would have been interesting to compare the extent of the first trajectory to the trajectory of the pre-trial movement in which the subjects actively defined the proprioceptive target. However, this was not possible because the optical capture system did not begin sampling until after the subjects selected the target. Consequently, the start position of the pre-trial trajectory could not be determined because it was not recorded. Given that the data suggest a relationship between movement extent and the trajectory start and endpoint positions, it is reasonable to suspect that the change in extent between the pre-trial and first movement repetitions would have been similar to the change in position between the target and the first endpoint position (i.e., the change in extent for the second repetition would have been less than the change in extent for the first repetition).

The present findings with respect to movement extent are particularly interesting because they conflict with the results of a similar study conducted by Brown *et al.* (2003a). In this study, subjects performed 75 non-visual, planar repetitive arm movements to visually defined targets from two different locations in the physical workspace. It was found that movement direction and extent were preserved, despite the fact that trajectory start positions drifted away from the original location (hand trajectory drift). Consequently, it was suggested that the processes needed to accurately reproduce limb movement were either unaffected by, or controlled separately from, the processes needed to accurately reproduce absolute spatial limb position. The results of this study provided support for what is known as the *separate controllers hypothesis*, which states that the CNS controls limb movement and limb position independently (Brown *et al.*, 2003a; Brown *et al.*, 2003b). This study also demonstrated that the preservation of

movement direction and extent occurred because joint torque profiles were modified to compensate for the effects of hand trajectory drift. Forward dynamics simulations revealed that without these joint torque modifications, the accumulated errors in spatial hand trajectory position would have resulted in substantial changes in movement direction.

With respect to the interpretation of the current results, it is conceivable that the location of the trajectory start and endpoint positions for each repetition were indirectly prescribed through the control of movement extent, such that the estimate of extent determined the relative distance between the two endpoints, and that this process of estimating extent improved after several repetitions of the trial. However, such a process does not provide an explanation for the accumulated deficits in spatial hand position error, or for the improved relative positioning of trajectory endpoints with respect to the previous repetition. Conversely, if the capacity to predict and reproduce the absolute three-dimensional spatial position of both trajectory endpoints improved, then decreased instantaneous differences in spatial position reproduction at both ends of the trajectory and simultaneous instantaneous improvements in movement extent would be expected. Consequently, with respect to the control of limb movement and limb position the results of this study would appear to suggest one of two possibilities: (1) if two separate controllers for limb movement and limb position exist, then both may have benefited from augmented motion-dependent proprioceptive feedback, such that the reproduction capability for absolute spatial position and movement extent improved, or (2) that only the capacity to estimate the absolute spatial positions of the trajectory endpoints improved, such that the reductions in the three-dimensional instantaneous start and

endpoint position errors were indirectly responsible for the instantaneous improvement in movement extent.

In conclusion, the similarity of the trajectory extent data to the start and endpoint position data suggests a similar process for the control of upper limb movement based on dynamic proprioception. For example, if the absolute values (positive instead of negative) are used to plot the extent data shown in Figure 3.5, the patterns of accumulated and instantaneous change are identical to those depicted for the trajectory start and endpoint positions shown in Figures 3.2 and 3.3. In this context, regardless of whether movement reproduction is defined in terms of the successive spatial positioning of trajectory endpoints or in terms of the reproduction of movement extent, or both, the current results demonstrate that movement reproduction capability improves immediately after the active definition of a proprioceptive target, as well as after the first movement reproduction. The present findings suggest that the pre-trial initial conditions, which were based on proprioceptive information about the target hand position, limb position, and the task specific limb dynamics, were modified and improved through a process in which dynamic proprioception was responsible for the refinement of motor output, possibly by providing updated position and/or limb dynamics information to a central internal model of the task.

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4

Dynamic Upper Limb Proprioception in Multidirectional Shoulder Instability

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**Published in Clinical Orthopaedics and Related Research 2004; 420: 181-189*

**This chapter is a slightly modified version of the paper cited above*

Abstract

The ability of subjects with multidirectional shoulder instability to use proprioception to complete a series of upper limb repositioning tasks was investigated. Twelve subjects with multidirectional instability and 12 control subjects were blindfolded and instructed to use proprioception to reproduce a self-selected target position as accurately as possible. Subjects completed ten repetitions for each limb using three distinct upper limb movements: overhead reaching, scapular plane pointing and humeral external rotation with abduction. A three-dimensional video motion analysis system tracked limb position and determined spatial hand position error. Subjects with multidirectional shoulder instability showed significantly greater hand position error than control subjects. No hand position error differences were found between the symptomatic and the asymptomatic limbs of subjects in the instability group. Inter-repetition error for subjects in both groups improved significantly during the first three movement cycles. The results suggested that after movement initiation, dynamic proprioception was a factor in improving hand position accuracy in both groups, but to a lesser degree in subjects with multidirectional instability. Consequently, subjects with multidirectional instability may have a reduced capacity to use proprioception to refine and control the motor output of the upper limb.

4.1 Introduction

Multidirectional instability of the shoulder is a complex condition that can be difficult to diagnose and to treat (An, 2000; Arendt, 1988; Lebar & Alexander, 1992; Mallon & Speer, 1995). Neer and Foster (1980) first recognized multidirectional instability as a unique and separate condition from unidirectional instability, and developed the inferior capsular shift as a specific surgical procedure for its treatment. Multidirectional instability can occur in both males and females, in different age groups, and in most segments of the population, from sedentary individuals to elite athletes (Arendt, 1988) and is considered to be a serious and more prevalent condition than previously realized (An, 2000). It is characterized by symptomatic global laxity of the glenohumeral joint (Beasley *et al.*, 2000) and may present either traumatically, atraumatically, unilaterally, bilaterally, or with or without generalized joint laxity (Brown *et al.*, 2000; Emery & Mullaji, 1991; Lebar & Alexander, 1992; Schenk & Brems, 1998). Individuals who possess multidirectional instability subluxate or dislocate anteriorly, posteriorly, or inferiorly with concurrent reproduction of symptoms in at least two directions (Beasley *et al.*, 2000; Schenk & Brems, 1998). Symptoms typically are associated with the midrange positions of glenohumeral motion and often occur during activities of daily living (Beasley *et al.*, 2000; Schenk & Brems, 1998).

The glenohumeral joint's relatively poor osseous and capsuloligamentous stability necessitates a reliance on neuromuscular proprioception and stabilization more than any other joint in the human body (Nyland *et al.*, 1998). Proprioception is essential for coordination of multijoint movement sequences (Bevan *et al.*, 1994; Cordo *et al.*, 1994;

Ghez *et al.*, 1990; Ghez & Sainburg, 1995), for regulating the amplitude and timing of antagonistic muscle activation for the purpose of limb segment deceleration (Forget & Lamarre, 1987), and for regulating the contraction of distal segment muscles to control proximally produced joint interaction torques (Sainburg *et al.*, 1993; Smith & Zernicke, 1987). Proprioceptive deficiencies, which exist in patients with functional deafferentation, create major deficits in movement control (Ghez *et al.*, 1995; Gordon *et al.*, 1994; Marsden *et al.*, 1984; Sainburg *et al.*, 1993; Sainburg *et al.*, 1995; Sanes *et al.*, 1985) which result in increased movement variability (Marsden *et al.*, 1984; Sanes *et al.*, 1985), the inability to maintain stable hand postures without visual guidance (Sanes *et al.*, 1985), and a reduced capacity to detect and correct motion errors based on limb movement information following the execution of steady-state motor commands (Ghez *et al.*, 1990; Gordon *et al.*, 1995; Marsden *et al.*, 1984; Sainburg *et al.*, 1993; Sanes *et al.*, 1985). Studies of the relationship between proprioception and joint stability acknowledge the complementary role of sensorimotor systems and passive joint structures to the maintenance of functional joint stability (Beard *et al.*, 2000; Iwasa *et al.*, 2000; Jerosch & Prymka, 1996). Proprioception provides a mechanism by which the neuromuscular control system operates as a kind of injury prophylactic to maintain the functional integrity and stability of the joint (Beard *et al.*, 2000; Iwasa *et al.*, 2000; Jerosch & Prymka, 1996). Muscle stiffness, muscle response time (latency), and passive and dynamic (active) joint repositioning tests have been used to quantify proprioceptive function (Beard *et al.*, 2000; Brindle *et al.*, 1999).

Given the importance of proprioceptive input to the neuromuscular control of limb movement and joint stability, it is surprising that no attempt has been made to investigate

the presence of motor performance deficits associated with upper limb proprioception in individuals with multidirectional instability. This is particularly remarkable given the recent interest in the contribution of proprioceptive feedback control mechanisms to functional joint stability in the knee and to injuries of the anterior cruciate ligament (Beard *et al.*, 2000; Beynnon *et al.*, 1999; Birmingham *et al.*, 1998; Birmingham *et al.*, 2000; Iwasa *et al.*, 2000; Jerosch & Prymka, 1996). Studies investigating proprioception and shoulder instability are limited and have utilized either passive (Blasier *et al.*, 1994; Lephart *et al.*, 1994; Voight *et al.*, 1996) or dynamic (Cohen, 1958; Forwell & Carnahan, 1996; Jerosch *et al.*, 1996) joint repositioning tests to assess proprioceptive function. To our knowledge, no study has investigated passive or dynamic proprioception in multidirectional shoulder instability. Therefore, the purpose of the current study was to use a dynamic, non-visually guided limb-repositioning test to investigate upper limb proprioception in subjects with multidirectional instability. Hand position errors of symptomatic and asymptomatic limbs in the instability group were compared with each other and with healthy control subjects' limbs. It was hypothesized that limbs with multidirectional shoulder instability would show greater hand positioning error when compared with asymptomatic limbs in the instability group and asymptomatic limbs in the control group.

4.2 Materials and Methods

Subjects

Twelve subjects with multidirectional instability (nine men, three women; age, 27.0 ± 8.3 years) and 12 control subjects (nine men, three women; age, 26.3 ± 5.4 years) with

normal, healthy shoulders participated in the study. Both shoulders were tested in all subjects. Subjects in the multidirectional instability group were tested after the original clinical diagnosis and did not receive any treatment or intervention before the test session. Pagnani and Warren (1994) and Brown *et al.* (2000) have classified multidirectional instability according to three subsets that include: (1) acute trauma, repetitive trauma or no trauma; (2) generalized joint laxity or isolated shoulder laxity; and (3) unilateral or bilateral symptoms. The 12 subjects with multidirectional instability tested in the current study were representative of all three subset categories. Of the 24 shoulders tested in the multidirectional instability group, 16 were symptomatic and eight were asymptomatic (four subjects were symptomatic bilaterally and eight unilaterally). Given that bilateral symptoms occur relatively frequently in multidirectional instability, it was not possible to test a homogeneous sample of unilateral subjects. Three of the eight subjects with unilateral instability had symptoms in the dominant limb, whereas five had symptoms in the nondominant limb. In addition, two of the subjects with bilateral instability were competitive athletes (one swimmer and one volleyball player). Patients were diagnosed and selected for inclusion to the multidirectional instability group based on the following criteria: (1) a functionally significant inability to keep the humeral head centered in the glenoid fossa, especially in positions not at the extremes of motion; (2) the absence of an injury mechanism likely to tear the glenohumeral ligaments; (3) spontaneous reductions of translations; (4) glenohumeral translations that duplicated the symptoms of concern to the patient; (5) a diminished resistance to translation in multiple directions as compared with a normal glenohumeral joint; and (6) an absence of traumatic lesions (Matsen III, 1994).

Exclusion criteria for subjects with multidirectional instability were mental incompetency, psychiatric or emotional difficulties related to voluntary instability and any musculoskeletal, neurologic, or genetic abnormality (such as Ehlers-Danlos syndrome) other than shoulder instability. Control subjects had no history of shoulder injuries, complaints or surgery and were selected to match the gender of subjects in the multidirectional instability group. Before participating in the study, subjects were required to indicate limb dominance and to provide individual informed consent. The current study was administered according to ethical guidelines and procedures outlined by our institution.

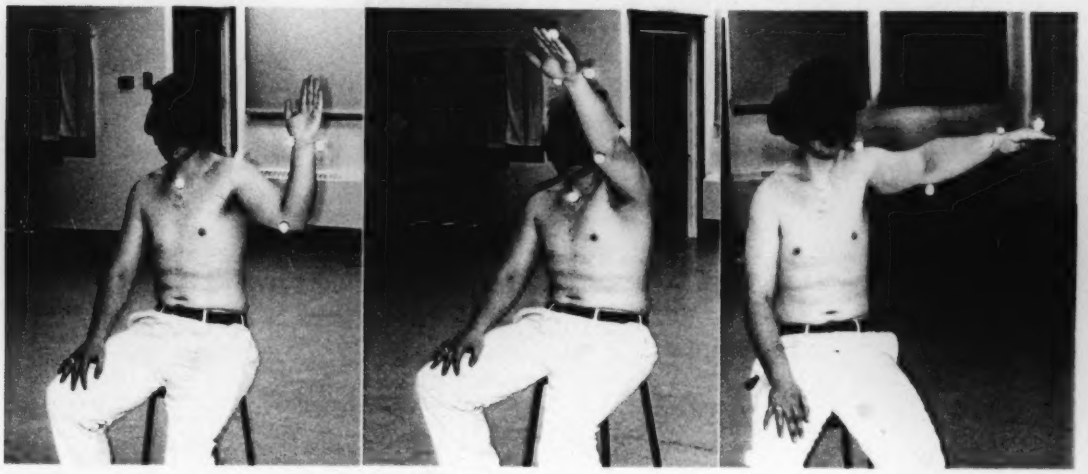
Movement and Test Protocol

Subjects were seated in the center of a rectangular test area bounded by four infrared video cameras. The cameras were arranged in an X-configuration to allow optimal three-dimensional reconstruction of non-invasive skin-mounted markers. The test required subjects to self-select a position in space with the hand (a virtual proprioceptive target) using one of three end point position limb configurations. The objective of the task was to move the upper limb in a specified manner and to return the hand to the target position as accurately as possible without visual guidance (subjects were blindfolded). Movements were initiated from the target position so subjects would use limb proprioception to perform the tasks.

A dynamic, rather than a passive, positioning test was chosen to measure proprioceptive function because passive tests are non-functional and less generalizable to the specific circumstances in which multidirectional instability occurs, those being

glenohumeral instability during active muscle contraction. Passive tests are potentially problematic because it can be difficult to control for differences in muscle spindle sensitivity (intrafusal muscle fiber slack) and cutaneous feedback from contact with the passive motion apparatus (Proske *et al.*, 2000). Evidence exists which supports the use of natural unrestricted arm movements as an effective strategy to gain insight into the planning and control processes of the central nervous system (Atkeson & Hollerbach, 1985).

Subjects repositioned the limb ten times for each of three upper limb movements: overhead reaching, scapular plane pointing and a combination of shoulder abduction with external rotation.



A) *B)* *C)*

Figure 4.1 A-C: *A blindfolded subject indicates the end point position for each of the three test movements: (A) combined shoulder external rotation and abduction, (B)*

overhead reaching, and (C) scapular plane pointing. White spheres indicate the retroreflective markers tracked by the motion analysis system.

These movements included the middle and end-ranges of shoulder motion and were selected to closely approximate movements associated with activities of daily living (for example, reaching for an object on a shelf). The rationale for assessing repetitive (cyclic) reaching movements in the current study was twofold. First, the coordination of cyclic movements has been shown to be critically dependent on proprioception (Cordo *et al.*, 1994). Second, it was important to have a sufficient number of repetitions to obtain a reasonable assessment of movement consistency. The end point position for shoulder abduction and external rotation (Figure 4.1A) consisted of the shoulder approaching full external rotation in 90° of abduction with the elbow flexed 90°. This movement was selected because, unlike the other tasks, it requires motion about the longitudinal axis of the humerus (internal and external rotation) and can be a challenging position for subjects with multidirectional instability. Overhead reaching (Figure 4.1B) consisted of raising the limb to an upright position above the head (oriented perpendicularly with respect to the horizontal plane). For scapular plane pointing (Figure 4.1C), the arm was elevated so that it was oriented in the scapular plane (approximately 30° anterior to the coronal plane), parallel to the horizontal and with the elbow in a slightly flexed position.

Movement demonstrations and verbal instructions concerning the specifics of the experiment were given to each subject at the beginning of the test session. Subjects were blindfolded and instructed to complete a minimum of three practice repetitions until they were comfortable with each of the test movements. These practice repetitions were done

immediately before the beginning of each movement trial. Subjects were instructed to hold the wrist steady, to use only the shoulder and elbow to return the hand to the target position 10 times, and to select a movement speed that would enable them to complete the task with the greatest accuracy. After the practice trials, subjects were instructed to place their limb in the appropriate end point position configuration and select the target position and, after a pause of several seconds, were given the command to start. For each movement the arm was lowered to a point on the ipsilateral thigh just below the hip (approximately 3/4 of the thigh length proximal to the knee) before it was returned to the target position using the movement specified for that trial. No performance feedback of any kind (knowledge of results or verbal encouragement) was given as to the outcome or accuracy of each repetition or trial. Subjects were instructed not to count, to complete all 10 repetitions of the trial without pause, and were told by the experimenter when to stop. Trial order (left or right arm and movement type) was randomized and all movements were tested on the same limb before testing the opposite side.

Apparatus and Measurement

Selected trunk and upper limb marker positions were sampled at 60 Hz using a three-dimensional video motion analysis system (Vicon Motion Systems, Lake Forest, CA). Nine retroreflective markers (1 cm in diameter) were secured to the following landmarks on each subject: the suprasternal notch, the ends of the spinous processes of C5 and T7, the lateral aspect of the acromion, the medial and lateral epicondyles of the humerus, the radial and ulnar styloid processes, and the middle metacarpophalangeal joint of the hand (Figure 4.1). These markers permitted three-dimensional movement of the upper limb to

be recorded and reconstructed without interruption. The complete marker set was used so that subsequent analyses could investigate the kinematics of the entire upper limb and trunk. Studies indicate that trajectories of goal-directed arm movements are planned in terms of hand location and subsequently transformed into the required joint rotations needed for movement execution (that is, the central nervous system uses a hand-centered coordinate system to plan arm movements) (Abend *et al.*, 1982; Flash & Hogan, 1985; Gordon *et al.*, 1994; Hollerbach & Flash, 1982; Lacquaniti & Soechting, 1982; Morasso, 1981; Wolpert *et al.*, 1995). Consequently, the current study used the three-dimensional position of the hand marker to quantify upper limb-positioning error.

Data Analysis

Three-dimensional hand marker position data were smoothed with a zero-lag, second order Butterworth low-pass digital filter (The Mathworks Inc., Natick, MA) at a cutoff frequency of 6 Hz (Winter, 1990). This was done to eliminate any subjectivity in identifying the appropriate video frame for the end point position of each repetition. Hand marker end point positions, which were used to calculate hand position error, were determined by visual approximation of the direction reversal point of the three-dimensional hand marker trajectory (Figure 4.2). The minimum tangential hand velocity for the reaching phase of each repetition was used to validate the frame number values determined by the hand trajectory method. For most trials, the frame numbers of the two end points were identical, and in the few cases where there was a discrepancy, differed by no more than two frames. The position differences in these cases were negligible (as the hand was stationary at this point) and the end point coordinates for the frame number

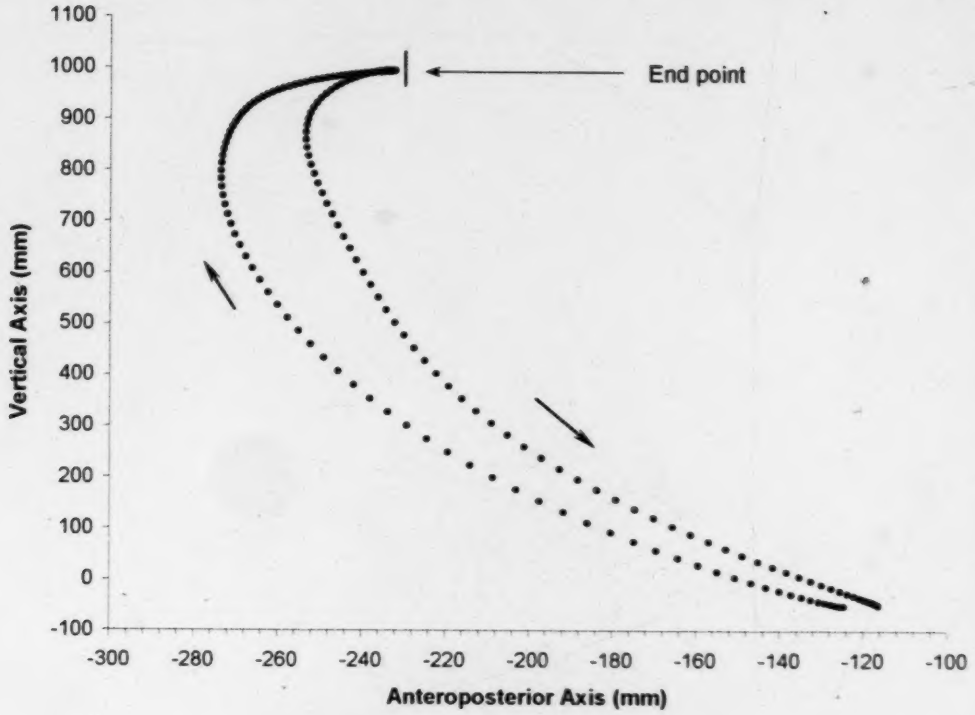


Figure 4.2: *A representative two-dimensional trajectory (one complete cycle for overhead reaching) indicates movement direction (arrows) and direction reversal at the end point position. Animation software (Vicon Corp.) permitted single-frame observation of the three-dimensional trajectory and was used to select the appropriate frame for the end point of each cycle. Three-dimensional marker coordinate positions for the end point frame were identified and recorded for each repetition.*

identified by the hand trajectory method were used for the position error calculations.

Dynamic three-dimensional radial hand position error was calculated so that the ability of each subject to consistently position the hand while moving the upper limb in a

specified manner, using only proprioception, could be determined. The method for calculating the mean hand position error for each trial was:

$$\text{Hand Position Error} = \left(\sum_{n=2}^{10} \sqrt{(X_n - X_{n-1})^2 + (Y_n - Y_{n-1})^2 + (Z_n - Z_{n-1})^2} \right) \div 9$$

where X, Y, and Z represent global coordinate positions in three-dimensional orthogonal space and n is the repetition or cycle number. This method of calculating three-dimensional hand position error provides a global error estimate in which the error term for each coordinate is represented equally in the total (the scale of each axis does not bias or weight the error determination). The initial target position of the hand was not included in the calculation of dynamic hand position error because of reported differences in static versus dynamic joint positioning accuracy (Nyland *et al.*, 1998). In addition to the trial errors, the mean hand position errors for each of the 10 cycles were calculated so that the progression of errors could be examined (for example, the error between cycles 9 and 10 could be compared to the error between cycles 2 and 3).

Three separate analysis of variance (ANOVA) tests were used to statistically compare hand position errors across several factors. The first ANOVA compared hand position error in the instability group according to shoulder condition (symptomatic or asymptomatic), limb dominance, and movement type (condition x dominance x movement type). The second ANOVA compared hand position error between the control and the instability group according to limb dominance and movement type (group x dominance x movement type). The third ANOVA compared each group's hand position error according to cycle number (group x cycle) to determine whether the cyclic nature of

the movements had any affect on hand positioning accuracy. Following a significant *F*-ratio, post-hoc analyses used the Bonferroni-Dunn statistical test. Probability values less than 0.05 were reported as statistically significant. For the post-hoc comparisons involving cycle number, an adjusted *p* value of 0.0014 was used to determine statistical significance.

4.3 Results

The hand position errors of limbs with symptomatic shoulders were compared with limbs with asymptomatic shoulders in the multidirectional instability group (Table 4.1). The within group ANOVA (condition x dominance x movement type) revealed no significant main effects for limb condition [F (1, 58) = 0.26, $p < 0.61$], limb dominance [F (1, 58) = 0.54, $p < 0.47$] or movement type [F (2, 58) = 0.08, $p < 0.93$]. There also were no significant interaction effects between any of the factors. Consequently, hand position errors for limbs with symptomatic shoulders were not different from hand position errors for limbs with asymptomatic shoulders. Given that the instability group analysis was unable to show a dependence of hand position error on shoulder condition (instability or no instability), the symptomatic and asymptomatic errors were combined and compared with the control group errors.

The between group ANOVA (group x dominance x movement type) revealed a significant main effect for group [F (1, 129) = 4.10, $p < 0.05$] with no significant main effects for limb dominance [F (1, 129) = 0.34, $p < 0.56$] or movement type [F (2, 129) = 0.71, $p < 0.49$]. There were no significant interaction effects between any of the three variables. The instability group mean hand position error was greater than the mean

Table 4.1: Control and Instability Group Hand Position Error (HPE) by Dominance and Condition (Mean ± 1 Standard Deviation)

	Control HPE (mm)	Instability HPE (mm)
Dominant	32.7 ± 9.3	37.3 ± 9.7
Nondominant	34.8 ± 9.4	37.2 ± 10.9
Symptomatic	–	37.5 ± 10.6
Asymptomatic	–	36.5 ± 9.9

control error for each of the three movement tasks (Figure 4.3). Therefore, subjects with multidirectional shoulder instability were unable to use proprioception to reposition the hand as accurately as control subjects, regardless of the type of movement used.

In addition to investigating the mean hand position errors for each trial, the individual repetition errors also were analyzed to investigate the cyclic nature of the movements and to obtain a more complete description of hand positioning error as the task progressed. This analysis was done to determine whether position error differences between the two groups depended on the cycle number or were consistent throughout the trial. A two-way ANOVA (group x cycle) was used to determine the effect of the repetition number on the hand position error of each group. Significant main effects were found for both group [F (1, 1401) = 7.43, $p < 0.01$] and cycle number [F (9, 1401) = 4.03, $p < 0.0001$]. The interaction between group type and cycle number was not significant. As with the mean hand position error results, this analysis indicated that subjects with multidirectional shoulder instability possessed greater hand positioning error than controls.

Figure 4.4 shows the individual position errors (cycle deviations) averaged across

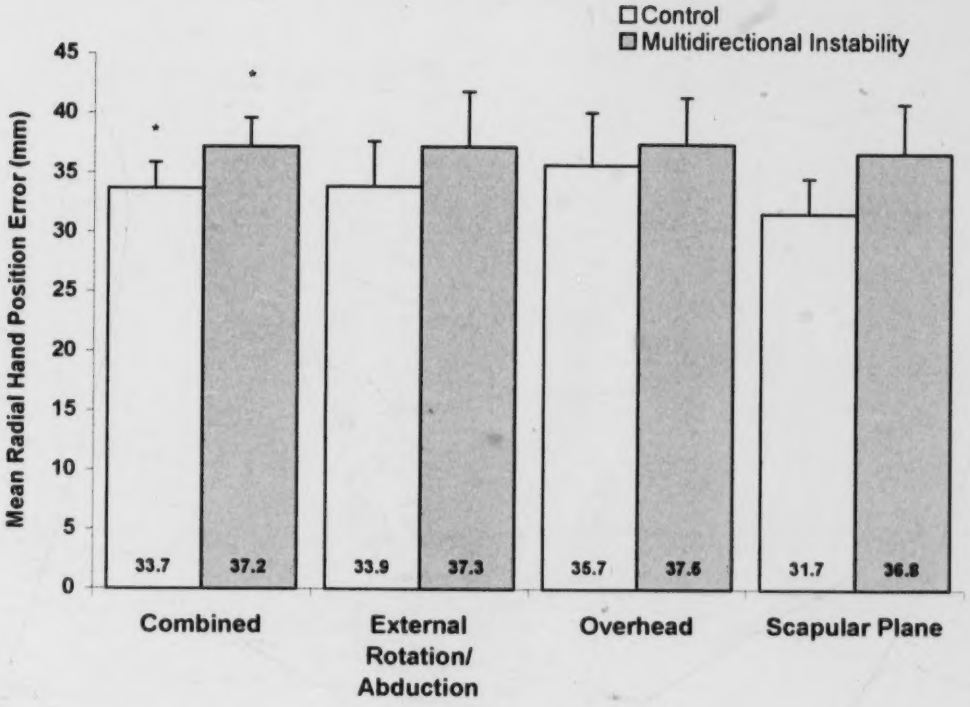


Figure 4.3: *Mean hand position error for control and multidirectional instability group subjects according to movement type indicates the instability group's hand position error was greater than the control group's error for each of the three test movements. The group main effect was significant (* = $p < 0.04$) with no interaction between group and movement type factors. There was no significant difference between movements ($p = 0.51$). The error bars show the 95% confidence interval.*

all trials for each group. The deviation between the first cycle and the starting position (the proprioceptive target), designated as 1-S, was not used in the calculation of mean hand position error (Figure 4.3) because it involves a static hand position. It is shown

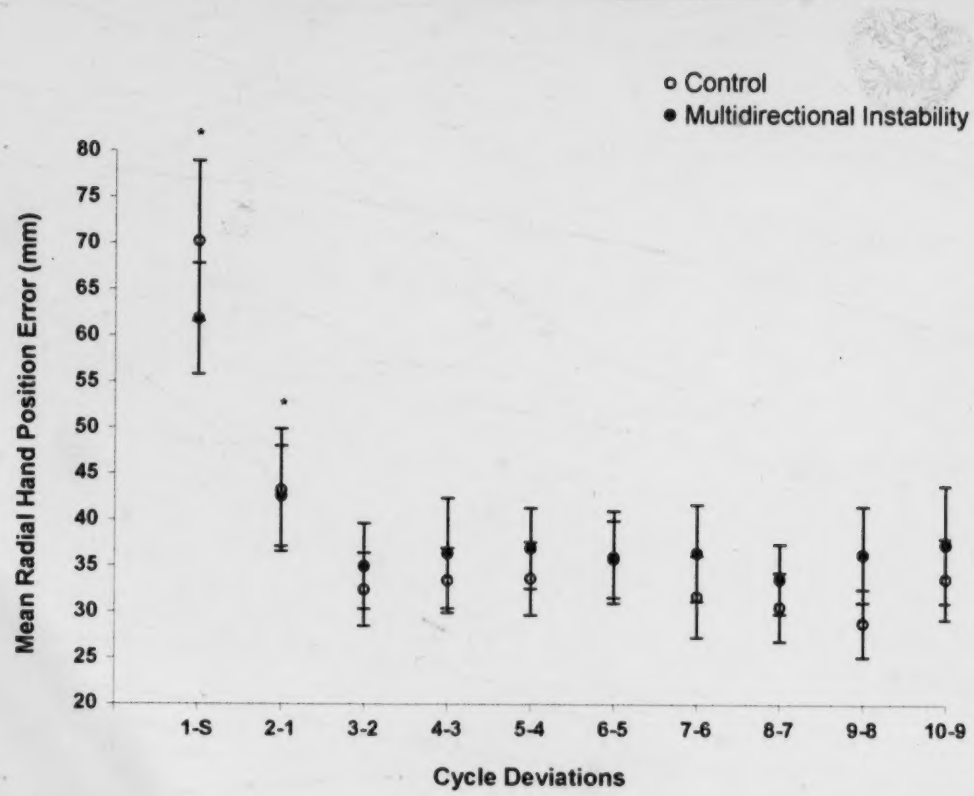


Figure 4.4: *The control and multidirectional instability groups' hand position errors for each of the 10 movement cycles were averaged across all trials. Each deviation indicates the n – (n – 1) cycle position error (S = start). The first cycle deviation (1-S) was not used to calculate hand position error, but is shown to illustrate the decrease in error that occurred immediately after movement initiation. Dynamic hand position errors (after movement initiation) for subjects with multidirectional instability were significantly greater than those of control subjects ($p < 0.01$). The first (1-S) and second (2–1) cycle deviations were significantly different from each other and from the other deviations of the trial (* = $p < 0.0014$), indicating that hand position error improved for both groups after the start of the trial, but to a greater degree for the control group. There were no*

other significant between cycle differences. The error bars indicate the 95% confidence interval.

with the other cycle deviations in Figure 4.4 to demonstrate the change in hand position error that occurred immediately after movement initiation. Post-hoc analysis, with an adjusted alpha level of 0.0014 to account for multiple comparisons (nine), showed that the first (1-S) and second cycle deviations (cycle 2 –1) were significantly different from each other ($p < 0.0014$) and from the other eight deviations of the trial ($p < 0.0014$). There were no other significant differences between any of the other cycle deviations. The presence of significant group and cycle differences, with no interaction between the two variables, indicated that subjects with multidirectional shoulder instability consistently repositioned the hand with greater error than controls. It also indicated that the motor output variability associated with proprioceptive hand positioning improved significantly for both groups during the first three repetitions of the trial. Figure 4.4 shows that, in terms of the pattern or progression of hand position errors, subjects with multidirectional instability completed the tasks in the same manner as controls, but with less accuracy.

4.4 Discussion

The current study used hand position error to evaluate the ability of subjects with multidirectional shoulder instability to use proprioception to dynamically reposition the upper limb. Of the two hypotheses tested in this study, the hypothesis that symptomatic

limbs would show greater hand positioning error than asymptomatic limbs within the multidirectional instability group was not supported. This result differs from a study of subjects with traumatic anterior instability, in which passive position and motion detection tests revealed that symptomatic shoulders were unable to discriminate joint position as accurately as asymptomatic shoulders (Lephart *et al.*, 1994). Methodological differences between the two studies (passive versus dynamic positioning) or the type of shoulder instability (multidirectional versus anterior instability) could account for the discrepancy. In the current study, four of the 12 subjects with multidirectional instability had bilateral symptoms. Consequently, the number of asymptomatic shoulders (eight) was less than it would have been had it been possible to test a homogeneous sample of subjects with unilateral symptoms. This result is, at best, only preliminary given that the comparison was based on an unequal number of symptomatic (16) and asymptomatic (8) limbs. Although the unequal distribution of cases had no affect on the homogeneity of variance, a post-hoc power analysis indicated that this comparison had only 7% power to identify a significant difference. Therefore, it is conceivable that the lack of a significant difference between limbs with symptomatic and asymptomatic shoulders was attributable to a Type II statistical error.

Conversely, the hypothesis that subjects with multidirectional instability would commit significantly greater hand positioning errors than control subjects with healthy shoulders was supported. The results showed that subjects with multidirectional instability were not able to use proprioception to control the upper limb and reposition the hand as accurately as control subjects. This reduced capability suggests that the proprioceptive input needed to regulate limb movement and maintain a functionally

stable glenohumeral joint in subjects with multidirectional instability may be deficient compared with control subjects with normal, healthy shoulders. Subjects with traumatic anterior shoulder instability (Lephart *et al.*, 1994) and clinically determined glenohumeral joint laxity (Blasier *et al.*, 1994) have been found to have decreased proprioception compared to subjects with normal shoulders. Other studies have shown that a complete lack of proprioceptive input (which causes functional deafferentation) creates difficulties in the ability to detect and correct motor output errors, which accumulate over successive repetitions to the point where the movement deteriorates and becomes unrecognizable (Forget & Lamarre, 1987; Ghez *et al.*, 1990; Ghez *et al.*, 1995; Gordon *et al.*, 1995; Marsden *et al.*, 1984; Sainburg *et al.*, 1993; Sainburg *et al.*, 1995; Sanes *et al.*, 1985). Although the level of hand position error was significantly higher for subjects with multidirectional instability, the cycle error results showed that no such process of error accumulation occurred, and the pattern of inter-repetition errors throughout the trial was similar for both groups. Consequently, subjects with multidirectional instability responded to the demands of the task in the same manner as control subjects, with both groups showing a significant improvement in positioning accuracy during the first three repetitions of the trial. The development of a theory to explain this improvement is crucial to understanding why subjects with multidirectional instability were unable to use proprioception to reposition the hand as accurately as control subjects.

The significant reduction in hand position error that occurred after movement initiation was most likely the result of the central nervous system not being able to obtain critical information about the initial configuration of the limb and the starting position of the hand (the target). Studies show that the central nervous system depends on visual

input for feed-forward control and the planning of movement trajectories (Ghez *et al.*, 1990; Polit & Bizzi, 1979; Sainburg *et al*, 1995), and that vision typically dominates over proprioception for target-pointing movements (Taylor & McCloskey, 1988). Polit and Bizzi (1979) found that initiation of purposeful movement is not prevented by the elimination of limb proprioceptive feedback, thereby emphasizing the importance of feed-forward control processes. Ghez *et al.* (1990) discovered that reaching accuracy improved when patients with functional deafferentation were allowed to view the limb in either a static or a dynamic state, indicating that feed-forward control is able to compensate for movement deficits attributable to inadequate proprioception (Ghez *et al.*, 1990; Sainburg *et al.*, 1993). Moreover, being able to view the static position of the hand before initiating a non-visually guided movement has also been found to improve reaching accuracy compared with non-viewing conditions, presumably by permitting better encoding of the initial conditions of the motor apparatus (Desmurget *et al.*, 1997). These and other results (Flanders & Soechting, 1990; Rosetti *et al.*, 1994) show that visual input is critical for optimizing feed-forward control mechanisms via the pre-movement encoding of limb and target positions.

It also has been established that static proprioception is unable to provide the central nervous system with an accurate assessment of initial limb orientation to attain a spatial target (Desmurget *et al.*, 1997). Adaptation of joint and cutaneous afferents responsible for signaling static position information can result in a poor approximation of the initial conditions of the limb (known as proprioceptive drift) (Gandevia, 1996; Paillard & Brouchon, 1968; Wann & Ibrahim, 1992). The procedure in the current study required subjects to self-select and initiate movement from a static starting position after an

interval of approximately 2 to 3 seconds, in which the limb was held motionless while awaiting the command to start. This interval was consistent and intentionally short so subjects would not become fatigued as a result of holding the limb against gravity. However, the active maintenance of limb position, in which minute repositioning of the limb occurs, has been shown to prevent proprioceptive drift (Wann & Ibrahim, 1992). Therefore, it seems unlikely that receptor accommodation would have had any effect on hand position error, because static limb position was actively maintained during the short interval preceding the trial. Consequently, the large position error at the end of the first movement repetition was likely the result of insufficient feed-forward control caused by a lack of visual input.

However, a lack of feed-forward control provides only a partial explanation for the improvement in hand positioning accuracy, because it fails to address the mechanism responsible for correcting this deficiency. Although static proprioceptive information about initial limb conditions may have been inadequate, after the limb was in motion the quality and quantity of proprioceptive input received by the central nervous system should have improved, because of increased activation of muscle spindle receptors transmitting muscle length change and rate of change information (Bevan *et al.*, 1994). In addition, the presence of isotonic muscle contractions would have contributed to an increased discharge of shoulder and elbow joint afferents (Gandevia, *et al.*, 1996).

The importance of proprioceptive feedback in regulating performance of learned motor patterns as limb movements progress and evolve is underscored by several studies (Birmingham *et al.*, 1998; Bock & Eckmiller, 1986; Polit & Bizzi, 1979; Smith & Zernicke, 1987). Cyclic movements in the cat hindlimb indicate that motion-dependent

feedback is used to monitor changing limb dynamics when joint interaction torques increase to substantial levels (Smith & Zernicke, 1987). Consequently, the central nervous system does not have to preplan all of the compensatory muscle actions to control the joint interaction forces that develop during multijoint movement. In reaching movements, it has been shown that the initial pre-programmed joint torques required to move the hand are unable to compensate fully for limb inertial anisotropy, and that limb dynamics information is updated on a continual basis via proprioception for the modification of subsequent movements (Ghez *et al.*, 1990). It would seem, therefore, that the central nervous system launches reaching movements by using rules that are somewhat inadequate for muscle activation, and subsequently steers the movement based on motion-generated (dynamic) proprioceptive feedback. For the current results, it is reasonable to propose that dynamic proprioceptive feedback was used to refine the movement according to limb dynamics information not available to the central nervous system before movement initiation. Consequently, the improvement in hand positioning accuracy which occurred for both groups during the first three movement cycles was likely the result of a transition from a state of incomplete feed-forward control, because of the absence of visual input and inadequate static proprioception, to a superior dynamic state based on enhanced proprioceptive feedback. After this transition, the results indicate that positioning accuracy for subjects with multidirectional instability failed to improve to the same level as that of control subjects. If motion generated proprioceptive feedback was responsible for substantially reducing position error in both groups at the beginning of the trial by refining motor output according to enhanced limb dynamics information,

then the results suggest that subjects with multidirectional instability may have a reduced capacity to generate or effectively utilize proprioceptive feedback for this purpose.

An important and related question concerning the presence of a proprioceptive deficiency in subjects with multidirectional instability is whether it is limited to the limb with the instability or applies to both limbs in patients with unilateral symptoms. If a deficiency is present in both limbs, as the results of the current study suggest, then multidirectional instability may be a secondary phenomenon related, in part, to deficient proprioception. If greater hand positioning error were to occur as a direct result of the instability, the same level of error should not be present in the unaffected shoulder. As previously discussed, the current study is not able to satisfactorily address this question because the lack of a significant difference between symptomatic and asymptomatic limbs of subjects in the instability group may have occurred because of a statistical error attributable to sampling limitations. The specific nature of the relationship between proprioception and multidirectional instability and the question of which is a primary or a secondary factor in the etiology of this condition, are important issues that require additional investigation.

In summary, subjects with multidirectional instability were unable to position the hand as accurately as control subjects when relying exclusively on proprioception to complete repetitive upper limb movements. The results imply that proprioceptive training exercises (Kiss *et al.*, 2001; Lephart & Henry, 1996; Swanik *et al.*, 2002) should be routinely incorporated into physical therapy programs (Arendt, 1988) for the treatment of multidirectional instability. Physical rehabilitation consisting of shoulder muscle strengthening and neuromuscular control exercises has been shown to successfully

relieve symptoms of pain and instability (An, 2000; Beasley *et al.*, 2000). The relatively high success rate of non-surgical intervention in multidirectional instability (Burkhead & Rockwood, 1992) indicates that improved neuromuscular control can compensate, to a large degree, for deficient passive anatomical restraint mechanisms. The results also suggest that surgical procedures, when attempting to restore functional stability, should attempt to maintain proprioception by avoiding mechanoreceptor dense regions of the glenohumeral joint such as the tendinous section of the supraspinatus insertion and the posteroinferior region of the rotator cuff (Minaki *et al.*, 1999). Future investigations, using methods similar to the current study, should examine a greater number of subjects and an equal proportion of symptomatic and asymptomatic limbs to better understand the nature of the relationship between dynamic proprioception and multidirectional shoulder instability.

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5

The effect of shoulder instability on repetitive reaching to remembered proprioceptive targets

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Revised and resubmitted to the Journal of Motor Behavior

Abstract

The purpose of this study was to investigate whether deficient proprioception is a factor related to multidirectional instability (MDI) of the shoulder. Twenty-two subjects (11 MDI and 11 control) performed three distinct repetitive reaching tasks to remembered proprioceptive targets, in which hand trajectory position error, movement duration and extent, and velocity profile kinematic parameters were analyzed. Unlike control subjects, subjects with MDI failed to modify the hand kinematics of repeated movements and demonstrated a lower level of hand repositioning consistency. A deficiency associated with the ability to effectively utilize proprioceptive input to refine and control the dynamics of goal-directed reaching could explain why subjects with MDI were unable to adequately adapt motor output in response to repeated task performance.

5.1 Introduction

Reaching movements allow primates to interact with their environment, and with this ability humans perform a variety of tasks such as feeding, lifting and communicating. Moving the arm to acquire a three-dimensional spatial target requires a complex control system that must plan, predict, execute, evaluate and adapt the motor response to meet the desired objective. A failure or deficiency in any one of these processes can lead to a disruption or degradation in the capacity to generate movement. Consequently, it is important to understand these processes so that new and effective treatment strategies can be developed for individuals afflicted with motor disorders such as Parkinson's disease (Adamovich, Berkenblit, Hening, Sage, & Poizner, 2001; Verheul & Geuze, 2004), cerebellar ataxia (Bastian, Zackowski, & Thach, 2000), sensory neuropathy (Gordon, Ghilardi, & Ghez, 1995; Sanes, Mauritz, Dalakas, & Evarts, 1985) and stroke (McCrea, Eng, & Hodgson, 2002).

An important emerging concept in the field of sensorimotor control is the idea that the central nervous system (CNS) uses internal models (forward and inverse) to predict and modify the consequences of motor actions (Davidson & Wolpert, 2003; Wolpert & Ghahramani, 2000). This conceptual framework suggests that internal forward models make predictions about the future sensory state of the system based on a copy of the outgoing motor commands (efference copy), on the sensory feedback received from peripheral receptors and on the context specific knowledge of the system dynamics (Webb, 2004; Wolpert & Flanagan, 2001; Wolpert & Ghahramani, 2000; Proske, 2005). The concept of efference copy was firmly established through the work of Von Holst and

Sperry (Wolpert & Flanagan, 2001), and recent evidence suggests that a pathway projecting from the superior colliculus to the frontal eye fields of the cerebral cortex, through the medio-dorsal thalamus, conveys corollary discharge information in non-human primates (Sommer and Wurtz, 2002). Numerous studies have also demonstrated the importance of proprioceptive input in controlling proximally produced joint interaction torques (Sainburg, Poizner, & Ghez, 1993; Sainburg, Ghilardi, Poizner, & Ghez, 1995; Smith & Zernicke, 1987), in providing assistance to passive joint structures to maintain joint stability (Jerosch & Prymka, 1996) and in regulating rhythmic movements in animals (Bässler, 1986; Conway, Hultborn, & Kiehn, 1987; Lennard, 1985; Wolf & Pearson, 1988) and in humans (Dietz, 2002).

Reaching movements are partially dependent on the coordinated activation of the scapulohumeral and scapulothoracic muscles to move and position the shoulder throughout its considerable range of motion. Consequently, the presence of shoulder pain and/or dysfunction has a potentially debilitating effect on the ability to perform a variety of everyday tasks. Glenohumeral instability is a common problem associated with a number of different shoulder injuries, and multidirectional instability of the shoulder (MDI) is a particularly complex form of instability that can occur traumatically, atraumatically, unilaterally, bilaterally and with or without generalized joint laxity (Brown, Tan, & Kirkley, 2000). Individuals with MDI experience subluxation or dislocation of the glenohumeral joint anteriorly, posteriorly or inferiorly with simultaneous reproduction of symptoms in at least two directions (Beasley, Faryniarz, & Hannafin, 2000). Symptoms typically occur in the midrange positions of glenohumeral motion and are therefore frequently associated with the activities of daily living (Mallon

& Speer, 1995; Nyland, Caborn, & Johnson, 1998; Schenk & Brems, 1998). Because MDI is atraumatic and associated with capsuloligamentous laxity and compromised dynamic stability (Arendt, 1988; Beasley *et al.*, 2000; Mallon & Speer, 1995), it has been suggested that deficient proprioception may be a contributing factor to the pathoetiology of MDI (Mallon & Speer, 1995; Schenk & Brems, 1998).

Studies of passive joint proprioceptive acuity have suggested that deficient proprioception is a factor related to joint instability at the knee (Beynnon *et al.*, 1999; Iwasa *et al.*, 2000), and its relationship to anterior cruciate ligament (ACL) injury has also received considerable attention (Beard, Dodd, & Simpson, 2000; Birmingham *et al.*, 1998; Birmingham, Inglis, Kramer, & Vandervoort, 2000). A number of similar studies have also suggested that deficient proprioception is related to the pathophysiology of anterior shoulder instability (Blasier, Carpenter, & Huston, 1994; Lephart, Warner, Borsa, & Fu, 1994; Warner, Lephart, & Fu, 1996). Given that MDI is associated with atypical patterns of shoulder muscle activation (Barden, Balyk, Raso, Moreau, & Bagnall, 2005a; Morris, Kemp, & Frostick, 2004) and that joint position sense in the upper limb has been shown to be related to the centrally produced sense of effort as well as to peripheral afferent input (Walsh, Hesse, Morgan, & Proske, 2004), the purpose of this study was to employ an active arm repositioning test to investigate whether subjects with MDI have a proprioceptive deficiency in the symptomatic shoulder. We hypothesized that deficient proprioception would affect the sensorimotor control of the upper limb, such that the ability to reach to three-dimensional spatial targets in the absence of vision would be compromised. To investigate this possibility, three repetitive reaching tasks to remembered proprioceptive targets (Barden, Balyk, Raso, Moreau, & Bagnall, 2005b;

Brown, Rosenbaum, & Sainburg, 2003) were used to compare the hand trajectory endpoint position errors and velocity profile kinematics of subjects with MDI to control subjects with normal, healthy shoulders.

5.2 Materials and Methods

Subjects

Eleven subjects with MDI (8 men, 3 women; age, 25.9 ± 7.7 years) and 11 control subjects (8 men, 3 women; age, 27.2 ± 6.1 years) with normal, healthy shoulders participated in the study. Fifteen of the 22 shoulders in the MDI group were symptomatic for MDI (7 subjects with unilateral and 4 subjects with bilateral MDI). Eleven of these 15 shoulders were selected for testing in the current experiment (one shoulder was randomly selected from each bilateral subject). The sample of 11 symptomatic MDI shoulders consisted of 4 dominant and 7 non-dominant limbs.

The diagnosis of MDI was made by a practicing orthopaedic shoulder specialist (R.B.) according to the following criteria: (1) a functionally significant inability to keep the humeral head centered in the glenoid fossa, especially in positions not at the extremes of motion; (2) the absence of an injury mechanism likely to tear the glenohumeral ligaments; (3) spontaneous reductions of translations (i.e., re-alignment of the joint); (4) glenohumeral translations that duplicated the symptoms of concern to the patient; (5) a diminished resistance to translation in multiple directions as compared with a normal glenohumeral joint; and (6) an absence of traumatic lesions (Matsen III, 1994). Subjects in the MDI group possessed no genetic or neurological abnormalities (such as Ehlers-

Danlos syndrome or syringomyelia) and received no treatment or intervention prior to testing. Control subjects had no history of shoulder injuries, complaints or surgery and were matched to subjects in the MDI group according to gender and limb dominance (4 dominant and 7 non-dominant limbs were randomly selected for testing from subjects in the control group). Subjects provided individual informed consent prior to participating in the study, which was conducted in accordance with guidelines and procedures outlined by the authors' institutional ethics review board.

Apparatus

Subjects sat in the centre of a rectangular test area bounded by four infrared video cameras that were placed at equidistant locations from the subject (one camera in each corner). Nine retroreflective markers (1 cm in diameter) were placed on the following anatomical landmarks of each subject: the suprasternal notch, the ends of the spinous processes of C5 and T7, the lateral aspect of the acromion, the medial and lateral epicondyles of the humerus, the radial and ulnar styloid processes, and the base of the metacarpophalangeal joint of the middle finger (Fig. 5.1A-C). In the current study, the three-dimensional position of the hand marker was analyzed to determine the spatial accuracy and movement kinematics of the endpoint of the effector (i.e., the arm). An optical capture motion analysis system (Vicon Motion Systems, Lake Forest, CA) tracked the three-dimensional position of the skin-mounted markers with a sampling frequency of 60 Hz.

Figure 5.1 A-C: *Endpoint positions for each of the three pointing tasks: (A) shoulder external rotation with abduction, (B) overhead reaching, and (C) scapular plane pointing. The white spheres indicate the retroreflective markers tracked by the optical capture system.*

Methods

Subjects in this study performed three distinct repetitive reaching movements to actively defined, remembered proprioceptive targets in different locations of the physical workspace. The movements were: (a) shoulder external rotation with abduction, (b) overhead reaching and (c) scapular plane pointing (Fig. 5.1A-C). Overhead reaching required the arm to be raised to an upright position above the head, with the arm approximately perpendicular to the horizontal plane. Scapular plane pointing consisted of elevating the arm in the scapular plane (approximately 30° anterior to the coronal plane) with the arm parallel to the horizontal plane. Shoulder abduction with external rotation involved simultaneous external rotation and abduction of the shoulder to a position approaching full external rotation and 90° of shoulder abduction. Unlike the other tasks,

the shoulder abduction/external rotation task was primarily a single-joint movement (the elbow was maintained in approximately 90° of flexion) that involved motion about the anteroposterior axis of the shoulder and the longitudinal axis of the humerus. This movement was selected because it presented a potentially challenging position for subjects with MDI (the clinical apprehension test position). Overhead reaching and scapular plane pointing were selected because of their familiarity (subjects did not have to learn any novel tasks or complex motor skills) and generalizability to the wide repertoire of three-dimensional reaching movements routinely performed in everyday life (reaching to grasp objects above and at various positions in front of and lateral to the body).

This study used the spatial position of the hand to quantify the motor behaviour of the arm because the objective of reaching is to move the arm to the appropriate location in three-dimensional space so that the hand can interact with the environment (McCrea *et al.*, 2002). Numerous studies have presented evidence which suggests that the CNS uses a hand-centred coordinate system for the planning and execution of reaching movements (Atkeson & Hollerbach, 1985; Flash & Hogan, 1985; Kaminski & Gentile, 1989; Wolpert, Ghahramani, & Jordan, 1995) and as a result, hand position errors (Adamovich, Berkenblit, Fookson, & Poizner, 1998; Darling & Miller, 1993; Flanders & Soechting, 1990; Soechting & Flanders, 1989) and velocity profile kinematics (Elliott, Carson, Goodman & Chua, 1991; Milner & Ijaz, 1990; Nagasaki, 1989; van der Meulen, Gooskens, Denier van der Gon, Gielen & Wilhelm, 1990; Zhang & Chaffin, 1999) have been routinely employed to investigate the neural control processes associated with the production of goal-directed reaching movements. In relation to testing the hypothesis of

deficient proprioception, the rationale for reaching to a remembered proprioceptive target (a virtual target that corresponds to a proprioceptively sensed limb and/or joint position) was so that the CNS would possess only a proprioceptive, and not a visual, estimate of the initial limb position and joint configuration (Adamovich, Berkenblit, Fookson, & Poizner, 1999; Barden *et al.*, 2005b). In the current experiment, the initial state of the system, which was prescribed through the process of actively defining the target, consisted of a proprioceptive estimate of the spatial limb position and joint configuration, an efference copy or template of the control signals that were used to generate the movement (Sommer & Wurtz, 2002; Wolpert & Flanagan, 2001), a movement specific estimate of the limb dynamics needed to perform the task as well as a proprioceptive estimate of the static forces and joint torques needed to counter the effect of gravity while maintaining the limb at the target position (Adamovich *et al.*, 1999; Barden *et al.*, 2005b; Walsh *et al.*, 2004).

Investigations of joint proprioceptive function have typically employed passive test procedures that measure either the joint motion detection or joint reproduction capability of subjects with injured or uninjured joints (Birmingham *et al.*, 2000; Lephart *et al.*, 1994; Risberg, Beynnon, Peura, & Uh, 1999). While these methods may be advantageous with respect to joint specificity, they are not representative of how a functioning neuromuscular control system operates (Beynnon, Good & Risberg, 2002). Studies show that active and passive movements stimulate synaptic neuronal activity in different parts of the brain (Weiller *et al.*, 1996), and that even relatively simple movements generate an extraordinary amount of sensory feedback, such that the CNS requires a gating strategy to deal with the large influx of sensory reafference (Apps, 1999; Chapman, 1994; Collins,

Cameron, Gillard, & Prochazka, 1998; Lemon, 1999). In conducting an investigation into discrepancies in the knee joint position sense values reported in the literature, Grob *et al.* (2002) recently demonstrated that five different passive knee motion detection and joint reproduction tests produced five different sets of results in the same subjects, none of which were significantly correlated with each other. Consequently, both the validity and generalizability of passive test procedures to the conditions in which natural movement occurs should be investigated further.

In the current study, natural three-dimensional multijoint movements were used to perform an active arm-repositioning test, so that the results would have the greatest generalizability to the actual conditions in which proprioceptive input mediates the neuromuscular control of the upper limb. An active proprioceptive positioning test was also selected because passive proprioceptive acuity has been shown to be least sensitive in the midrange positions of glenohumeral motion (Blasier *et al.*, 1994; Janwantanakul, Magarey, Jones, & Dansie, 2001), and midrange instability is one of the characteristic features of MDI (An & Friedman, 2000; Arendt, 1988; Beasley *et al.*, 2000; Mallon & Speer, 1995; Schenk & Brems, 1998).

Protocol

Movement demonstrations and verbal instructions concerning the specifics of the protocol were given to each subject at the beginning of the test session. Subjects were blindfolded and instructed to select a movement speed that would enable them to reproduce the self-selected target position with the greatest accuracy. After the subjects were blindfolded, several practice repetitions (a minimum of 3) were performed at the

beginning of each trial, so the subjects would be comfortable and familiar with each task. After completing the practice repetitions, subjects moved the arm to the appropriate workspace location for that trial (overhead, scapular plane or external rotation/abduction) and a virtual target position was selected. The arm was maintained in this position for approximately 2 to 3 seconds until the subject was instructed to begin the trial. Once the start command was given, the arm was lowered to a position along the midline of the ipsilateral thigh just below the hip before being returned to the target. No extrinsic feedback of any kind (verbal encouragement or knowledge of performance) was given concerning the outcome or accuracy of each movement repetition or trial. Each subject performed three randomly assigned trials, one for each task. For each trial, the task required returning the hand to the proprioceptively sensed target position 10 times in succession. Consequently, each subject performed a total of 30 repetitive movements (3 tasks x 10 repetitions per trial). Subjects were instructed not to count (they were told by the experimenter when to stop), to maintain a fixed position at the wrist and to complete all ten repetitions of the trial consecutively.

Data Analysis

The three-dimensional hand marker position data were smoothed with a zero-lag, 2nd order Butterworth low-pass digital filter (The Mathworks Inc., Natick, MA) at a cutoff frequency of 6 Hz (Winter, 1990). The trajectory start and endpoint positions for each repetition were determined by viewing a three-dimensional representation of the trajectory to obtain a visual estimate of the direction reversal points. The three-dimensional coordinates of the appropriate video fields that corresponded to the

identified endpoints were then used to calculate the spatial parameters for that repetition. To corroborate the positions identified by the trajectory reversal method, endpoint positions using the minima of the three-dimensional tangential hand velocity profiles were also obtained. No discrepancies were found between the two methods.

For each trajectory, the spatial position error of the trajectory endpoint and the movement amplitude (extent) were calculated. The three-dimensional radial distance vector of each trajectory was used to determine movement extent. The difference in magnitude between each three-dimensional radial distance vector and the first trajectory vector was used to determine the accumulated change in movement extent. Hand trajectory endpoint position error was assessed using two aggregate (*x*, *y* and *z* directions combined) three-dimensional error parameters, the target position error (TPE) and the inter-repetition error (IRE). The target position error quantified the difference between each trajectory endpoint position and the actively defined proprioceptive target, whereas the IRE quantified the difference between each endpoint position and the previous endpoint position. A more detailed description of these error calculation methods can be found in a related publication by Barden *et al.* (2005b). In addition to quantifying the spatial parameters of the tasks, several kinematic variables based on the tangential hand velocity profiles were also determined. The tangential hand velocity profile refers to the time course of the resultant three-dimensional velocity vector for the reaching phase of each repetition (Adamovich *et al.*, 1998; Milner & Ijaz, 1990). To quantitatively compare the kinematics of reaching between the two groups, four hand velocity profile kinematic parameters (Fig. 5.2) were obtained: the peak velocity (PV), the average acceleration (AA), the average deceleration (AD) and the movement time (MT).

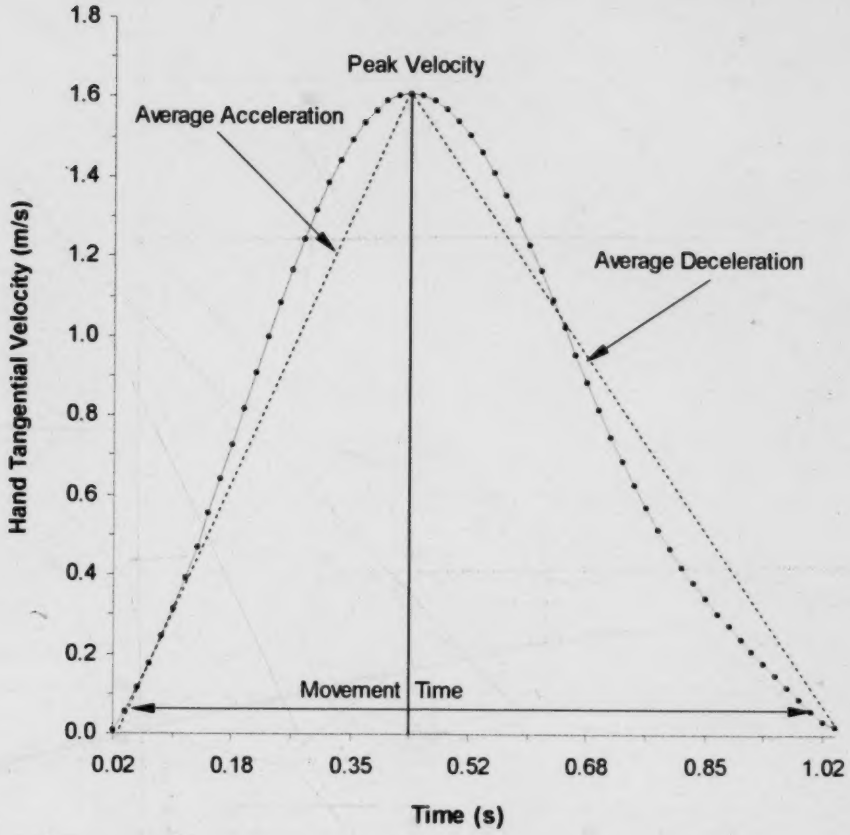

Figure 5.2: *Representative three-dimensional tangential hand velocity profile and associated kinematic parameters.*

The position error, movement extent and kinematic data were statistically analyzed using a one-between (group), two-within (repetition x task) three-way repeated measures analysis of variance (ANOVA) for each dependent variable. Following a significant *F*-ratio ($p < 0.05$), Bonferroni adjusted paired t-test comparisons were conducted for the 1st and 10th and the 1st and 2nd repetitions. Greenhouse-Geisser corrected degrees of freedom

were reported for cases in which the assumption of sphericity was not satisfied (Darlington, 1990).

5.3 Results

Hand position errors and movement extent

The data show that hand position error changed in response to repeated task performance, particularly at the beginning of the trial. For the target position error (TPE), which measured the hand trajectory endpoint position relative to the proprioceptive target, there was a significant within-subject main effect for repetition, F (4.07, 81.39) = 10.76, $p < 0.0001$, such that TPE accumulated for both groups over the course of the trial (Fig. 5.3A). Post hoc comparisons for repetitions 1 vs. 10 were significant for both groups ($p < 0.0001$). No other significant between or within-subjects effects were found for TPE. In contrast to TPE, the hand inter-repetition error (IRE) decreased immediately after the first repetition ($p < 0.01$ for both groups) and to a lesser extent throughout the rest of the trial (Fig. 5.3A). Consequently, the within-subject main effect for repetition was significant, F (4.01, 80.12) = 8.89, $p < 0.0001$, as were the post hoc comparisons for IREs 1 vs. 2 and 1 vs. 10 for both groups. In addition, a significant between-subject main effect for group was found, F (1, 20) = 5.63, $p < 0.05$, such that the IRE of the control group was significantly less than that of the MDI group. The results indicate that subjects with MDI improved the repeatability of their movements, but to a lesser extent than subjects with normal, healthy shoulders. No other significant main or interaction effects were found for IRE.

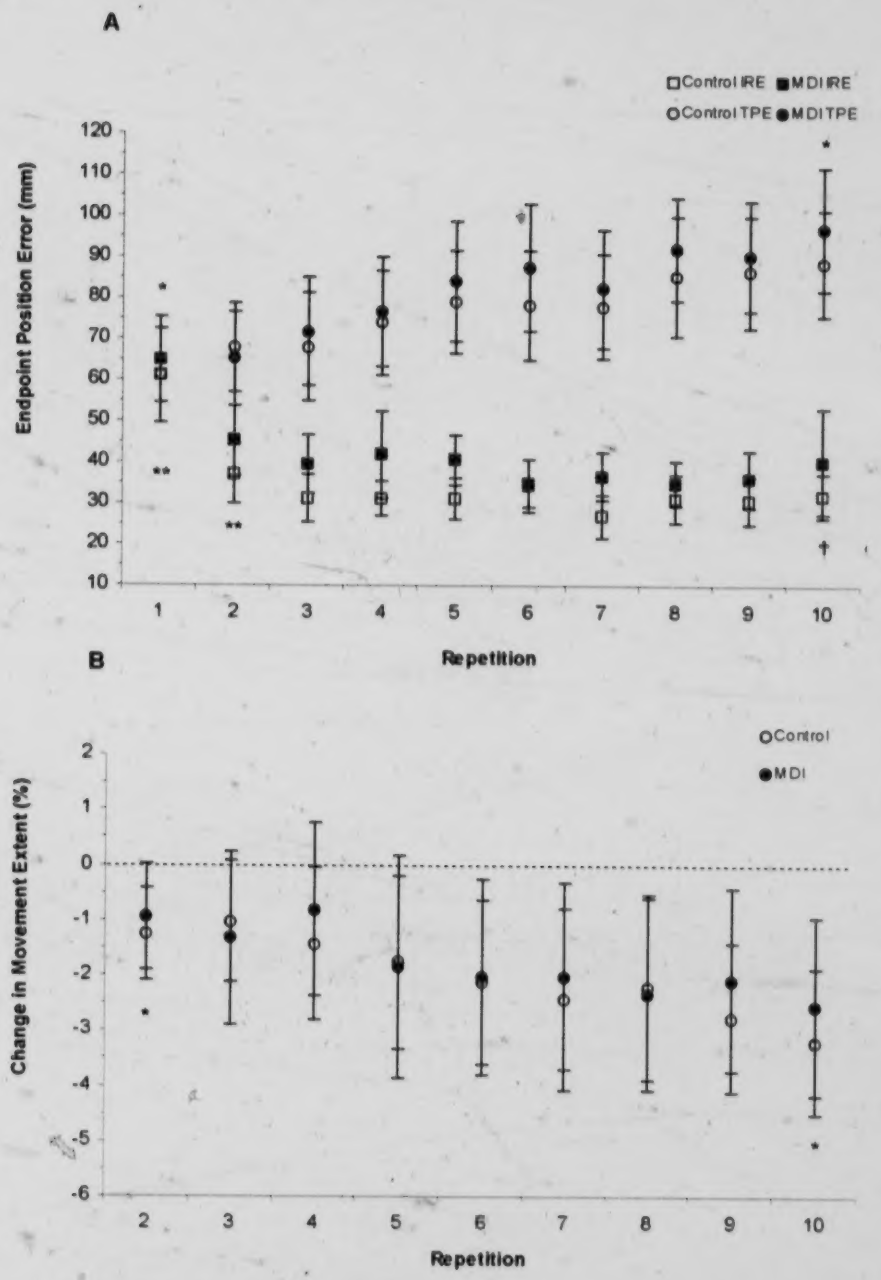

Figure 5.3: *Endpoint position error (A) and percent change in movement extent (B) for control and MDI subjects across all three tasks. TPE = target position error, IRE =*

inter-repetition error. Panel (A) demonstrates that target position error increased (= $p < 0.0001$) while endpoint inter-repetition error decreased († = $p < 0.0001$ for repetitions 1 vs. 10 and ** = $p < 0.01$ for repetitions 1 vs. 2) for both groups over the course of the trial. Control IRE was also significantly less than MDI IRE ($p < 0.05$). Panel (B) demonstrates that both groups consistently underestimated the extent of their movements (* = $p < 0.001$ and $p < 0.05$ for control and MDI subjects, respectively). The dashed line indicates movement extent for the first repetition. Error bars = 95% confidence interval.*

For movement extent, a significant within-subject main effect for repetition was found, F (3.26, 65.26) = 5.71, $p < 0.001$, indicating that subjects in both groups consistently underestimated the extent of their movements as the trial progressed (Fig. 5.3B). The progressive nature of the underestimation was confirmed by post hoc testing which revealed that the percent change in extent for repetition 10 was significantly greater than that for repetition 2 ($p < 0.001$ and $p < 0.05$ for control and MDI subjects, respectively). No other significant main or interaction effects were found for movement extent.

Velocity profile kinematics

The results for the four kinematic parameters, peak velocity (PV), movement time (MT), average acceleration (AA) and average deceleration (AD) were characterized by significant repetition x group interactions. For PV, a significant within-subject main effect for repetition was found, F (3.24, 64.78) = 6.52, $p < 0.0001$, in addition to a significant repetition x group interaction, F (3.24, 64.78) = 2.75, $p < 0.05$. Although not

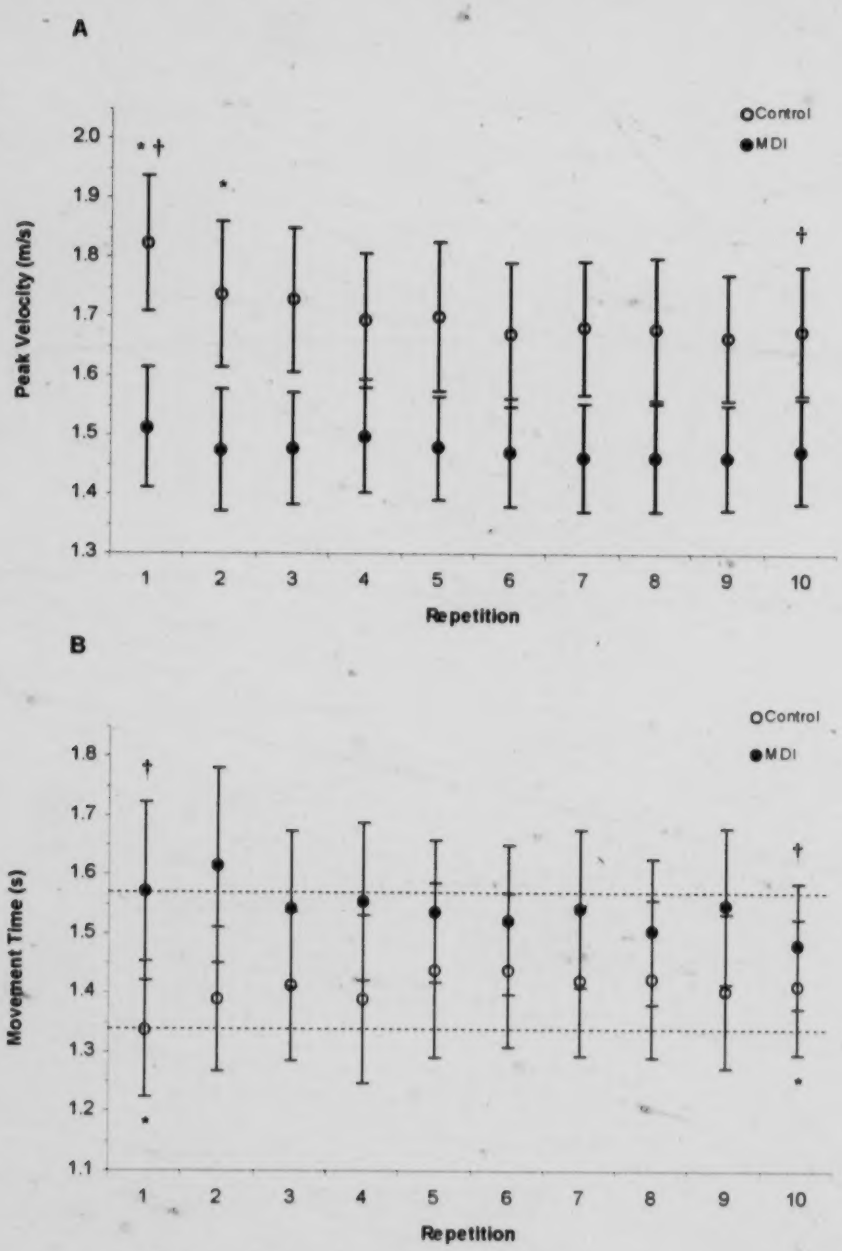

Figure 5.4: *Peak velocity (A) and movement time (B) for control and MDI subjects across all three tasks. Panel (A) shows that control subjects decreased the velocity of*

*their movements, *† = $p < 0.001$ for 1 vs. 2 (*) and 1 vs. 10 (†), whereas MDI subjects did not. In panel (B), movement time increased over the course of the trial for the control group (* = $p < 0.01$), whereas it decreased for subjects with MDI († = $p < 0.05$). The dashed lines provide a reference for the mean change in movement time relative to the first repetition of the trial. Error bars = 95% confidence interval.*

significant, the between-subject main effect for group approached significance at $p < 0.08$ (F (1, 20) = 3.28, $p < 0.08$). There were no other significant main or interaction effects for peak velocity. The data for PV are shown in Figure 5.4A, which demonstrates that PV for the control group decreased significantly after the first repetition ($p < 0.001$) and was then maintained throughout the rest of the trial ($p < 0.001$ for repetition 1 vs. 10). Post hoc comparisons for repetitions 1 vs. 2 and 1 vs. 10 confirmed that PV for the MDI group remained unchanged throughout the trial (the difference in mean PV between the two groups decreased by 35% between repetitions 1 and 10). Consequently, the results demonstrate that control subjects with healthy shoulders decreased the velocity of their movements with repeated task performance, whereas the subjects with MDI did not.

For movement time (MT), a significant repetition x group interaction was found, F (3.65, 76.63) = 3.25, $p < 0.01$, such that MT for the control group increased over the course of the trial, whereas it decreased for the MDI group. Post hoc comparisons confirmed that repetitions 1 vs. 10 were significantly different for both groups ($p < 0.01$ and $p < 0.05$ for control and MDI groups, respectively). There were no other significant within or between-subject effects for this variable. The mean movement time data for each repetition of the 10- repetition trial are presented in Figure 5.4B. The results for the

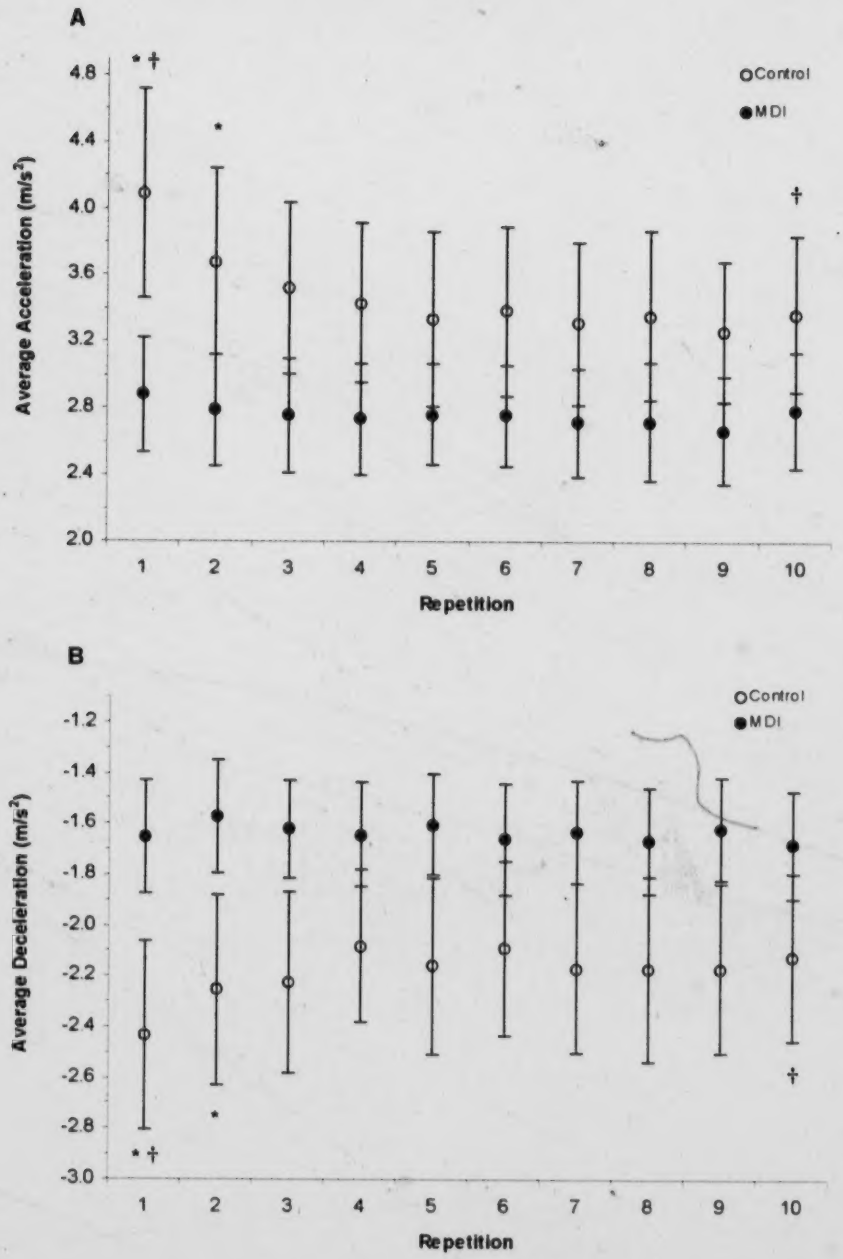

Figure 5.5: *Average acceleration (A) and deceleration (B) for control and MDI subjects across all three tasks. As with peak velocity, control subjects decreased the*

*average acceleration (A) and deceleration (B) of their movements, *† = $p < 0.001$ for 1 vs. 2 (*) and 1 vs. 10 (†), whereas MDI subjects did not. Error bars = 95% confidence interval.*

average acceleration (AA) and average deceleration (AD) were similar to those for PV. For AA, a significant within-subject main effect for repetition was found, F (2.77, 55.42) = 7.97, $p < 0.0001$, in addition to a significant repetition x group interaction, F (2.77, 55.42) = 3.78, $p < 0.01$. Similarly, for AD the within-subject main effect for repetition was significant, F (3.41, 68.21) = 2.04, $p < 0.05$, as was the repetition x group interaction, F (3.41, 68.21) = 3.28, $p < 0.001$. No other significant main or interaction effects were found for either AA or AD.

The data show that control subjects decreased the average acceleration of their movements over the course of the trial, but more so after the first and second repetitions (Fig. 5.5A). Post hoc analyses confirmed that control AA for repetitions 2 and 10 decreased significantly relative to the first repetition ($p < 0.001$). Unlike subjects in the control group, subjects with MDI made no modifications to the acceleration of their movements at any point during the trial. Similarly, the data indicate that control subjects decreased the deceleration of their movements while subjects in the MDI group did not (Fig. 5.5B). As with AA and PV, this decrease was more pronounced after the first and second repetitions than at the end of the trial. Post hoc comparisons for AD for repetitions 1 vs. 2 and 1 vs. 10 in the control group were significant ($p < 0.001$), whereas those in the MDI group were not. Generally speaking, the kinematic parameter results demonstrate that subjects with MDI were consistent in the production of their movements as

successive repetitions of the task were completed, but that this consistency was distinct from adjustments made by subjects in the control group that affected the motion of the hand.

5.4 Discussion

The purpose of the current study was to use several repetitive, three-dimensional proprioceptive reaching tasks to investigate whether subjects with MDI have a proprioceptive deficiency in the symptomatic shoulder. The hypothesis that subjects with MDI would demonstrate a compromised ability to reach to remembered proprioceptive targets was supported by the hand trajectory endpoint position error and movement kinematic results. Specifically, subjects with MDI demonstrated less precision in their ability to consistently reposition the hand relative to the previous movement, and failed to reproduce the hand velocity profile kinematic modifications that occurred in the control group. The following discussion of the hand trajectory endpoint position error, movement extent and velocity profile kinematic parameter results is presented in the context of current theories for the sensorimotor control of movement and recent findings for the performance of proprioceptive reaching tasks. While the current results support the hypothesis that subjects with MDI have a proprioceptive deficiency in the symptomatic shoulder, several alternative hypotheses to this explanation are also presented.

Hand position errors and movement extent

For subjects in both groups, the results demonstrate that the target position errors increased significantly over the course of the trial. The accumulated target position error

indicates that the hand trajectory endpoint position drifted relative to the target as further repetitions of the task were completed, and is consistent with recent findings of hand trajectory drift for repetitive three-dimensional (Barden *et al.*, 2005b) and planar proprioceptive reaching tasks (Brown *et al.*, 2003). Using the same methodology as that reported here, it has been shown that repetitive hand position errors relative to the initial position accumulate at both ends of the trajectory, indicating that hand trajectory position drifts over time when reaching to remembered proprioceptive targets (Barden *et al.*, 2005b). In the current study, subjects in both groups also consistently underestimated the extent of their movements for all three tasks, such that by the end of the trial, the average trajectory length had decreased by approximately 3% relative to the length of the first trajectory. These findings were also recently reported in healthy participants (Barden *et al.*, 2005b), but conflict with the results of Brown *et al.* (2003), who found that movement extent was maintained when consecutive planar, proprioceptive reaching movements were made to visually defined targets from two different locations in the physical workspace. The discrepancies in the results most likely occurred because of methodological differences related to the type of arm movements studied (two vs. three-dimensional) and/or the procedures used to define the targets (visual vs. proprioceptive).

Although the extent and endpoint positions of the trajectories relative to the target were similar for both groups, the inter-repetition errors (IRE) of the MDI group were significantly greater than those for the control group. This finding demonstrates that subjects with MDI were unable to repeatedly position the limb in three-dimensional space as precisely as control subjects with normal, healthy shoulders. Subjects in both groups showed a significant reduction in IRE after the second repetition, such that the inter-

repetition error between positions two and one was less than the error between position one and the target. This reduced IRE was then maintained throughout the rest of the trial in both groups, however the reduction was significantly less for the subjects with MDI. To explain this result, it is necessary to consider several important concepts that have emerged from the computational study of motor control (Wolpert & Ghahramani, 2000). One of these concepts is that the CNS uses internal models to plan, predict and control the large repertoire of human motor behaviour. Numerous studies have presented evidence in support of this concept (Barden *et al.*, 2005b; Beer, Dewald, & Rymer, 1999; Davidson & Wolpert, 2003; Flanagan, Vetter, Johansson, & Wolpert, 2003; Hinder & Milner, 2003; McIntyre, Zago, Bethoz, & Lacquaniti, 2001; Miall & Reckess, 2002; Miall, Weir, Wolpert, & Stein, 1993), which forms part of a larger conceptual framework for understanding the neural processes involved in the control of movement.

According to this framework, it is reasonable to suggest that the reduction in IRE occurred because there was a transition in limb states, namely the transition from a discrete, static proprioceptive state to a continuous, dynamic state that was sustained throughout the trial. Recent theories of motor learning and prediction (Davidson & Wolpert, 2003; Flanagan *et al.*, 2003; Wolpert & Ghahramani, 2000; Wolpert & Flanagan, 2001) suggest that the CNS uses an internal forward model to predict the future sensory state of the system based on the existing sensory information and an efference copy of the outgoing motor commands. The efference copy, along with information about the dynamics and current state of the system, generates a sensory prediction to which the actual sensory input resulting from the movement (the reafference) can be compared. An error signal is produced if the reafference differs from the prediction, which will

ultimately be used by an inverse dynamics model to generate new motor commands and subsequent updated predictions (Wolpert & Ghahramani, 2000). Interestingly, several studies have recently shown that the posterior parietal cortex is involved in the comparison of sensory and motor information during self-generated movements (MacDonald & Paus, 2003; Sirigu, Daprati, Pradat-Diehl, Franck, & Jeannerod, 1999), which suggests that it may function as the feedback controller from which the required motor error signal is produced (Wolpert & Ghahramani, 2000).

Consequently, when subjects in the current study actively defined the target, an efference copy and the available proprioceptive input (static and dynamic) would have been used to generate a prediction of the expected sensory feedback. Given the limb's initial state (i.e., the state which immediately preceded the command to begin the trial), this prediction would not have contained information about the lowering phase of the trajectory or of the specific limb dynamics associated with performing trajectory reversals (Sainburg *et al.*, 1993; Sainburg *et al.*, 1995). As a result, the actual sensory reafference (dynamic proprioception) associated with repetition #1 would have differed from the prediction, producing an error signal that was subsequently used to modify the current state estimate and generate a new prediction. A particularly well presented and detailed account of the process of state estimation and prediction has been presented by Wolpert & Ghahramani (2000), but for the purpose of the current discussion it is sufficient to state that the generation of a new prediction by the internal forward model presupposes a modification or correction to the motor commands based on the error signal from the feedback controller and the output of the inverse model.

With respect to the current data, it is reasonable to suggest that the reduction in IRE was related to the ability of the CNS to update an inverse dynamics model based on the actual and expected sensory feedback (Flanagan *et al.*, 2003). That is to say, the IRE reflects the degree of precision with which the CNS was able to reproduce the specific task given the prescribed set of experimental conditions. Therefore, the significant difference in IRE between the two groups suggests that some aspect of the sensorimotor control system is adversely affected in subjects with MDI. An explanation that is consistent with the hypothesis of this study is that subjects with MDI were not able to reduce IRE to the same level as control subjects because of a proprioceptive deficiency in the symptomatic shoulder. A proprioceptive deficiency would have a direct effect on the sensory reafference, which would compromise the ability to update the inverse dynamics model, which in turn would limit the fidelity of future state predictions. Consequently, it is reasonable to infer that a lower level of precision for motor prediction would equate to a lower level of precision for movement reproduction. Although theoretically possible, it seems less likely that the difference in IRE between the two groups would be related to a problem in the ability of subjects with MDI to produce an efference copy or with the ability of the internal model to make accurate predictions about the future sensory state of the system, given the same feedback as the subjects with healthy shoulders. Given the excessive glenohumeral laxity that is typically present in MDI, it is preferable to propose a mechanism whereby a lack of capsuloligamentous tension precipitates a decrease in the stimulation of local proprioceptive afferents. Theoretically, this deficiency in joint proprioceptive feedback exacerbates the existing passive insufficiency by compromising the neuromuscular control of the joint (Lephart & Henry, 1996).

The internal model interpretation of the data also provides a potential explanation for the accumulated increase in TPE, because it has been suggested that predictions made by internal forward models will drift over time if the models are not perfectly accurate (Wolpert & Flanagan, 2001). Given the current conditions (i.e., no vision), it is reasonable to expect such accumulated errors because the forward model would not have had any external reference or cue with which to update its predictions about the location of the arm in three-dimensional space. This explanation might also account for the accumulated deficits in movement extent, which in combination with the hand trajectory drift results suggests that the estimation of movement amplitude is related to the ability to predict and reproduce the absolute three-dimensional location of both trajectory endpoints (Barden *et al.*, 2005b). Generally speaking, the results imply that errors related to limb position and movement extent accumulate when the CNS lacks the appropriate external reference with which to calibrate its prediction. This interpretation is consistent with the recent findings of Barden *et al.* (2005b) and Brown *et al.* (2003), who demonstrated that, under similar conditions, movement extent was maintained for visually defined targets (Brown *et al.*, 2003) but not for targets that were defined exclusively with proprioceptive information (Barden *et al.*, 2005b).

Velocity profile kinematics

The tangential hand velocity profile kinematic results demonstrate that while subjects with normal shoulders adapted the kinematics of reaching with repeated task performance, the subjects with MDI did not. Specifically, the control group results showed that the first movement reproduction (i.e., the 2nd repetition) was associated with a decreased peak velocity, average acceleration and average deceleration for all three

tasks, which was then maintained, with minor modifications, throughout the rest of the trial. Similar reductions in the hand velocity profile kinematics of subjects with MDI did not occur. The results also showed that subjects in both groups altered the duration of their movements, but in significantly different ways, with control subjects increasing and MDI subjects decreasing movement duration over the course of the trial.

With respect to the source of the kinematic modifications in the control group, it has been shown previously that the CNS adjusts movement amplitude by scaling either the magnitude or duration of the accompanying velocity profile, and that the preferred strategy is to scale peak velocity rather than movement duration (Milner, 1986). Based on this relationship it is conceivable that the decreased velocity for the control group was related to the underestimation of movement extent, such that an intermediate strategy was adopted whereby both peak velocity and movement duration were scaled. However, subjects in the MDI group also underestimated movement extent, but demonstrated no changes in hand velocity profile kinematics other than a decrease in movement duration. Another possibility, which is consistent with the internal model explanation for the hand trajectory position error results, is that the reduction in control group velocity and acceleration for the second repetition occurred in relation to the modified state estimate and the subsequent new predictions generated by the internal forward model, which is to say that the modifications in hand kinematics were related to the reduction in IRE. Because the significant reductions in IRE for both groups indirectly suggests that the motor commands of subsequent movements were adapted in some way, the lack of modifications to the hand velocity profile kinematics of subjects with MDI suggests that some other difficulty, such as a problem with the interjoint coordination of the shoulder

and elbow and/or an inability to adequately compensate for the joint interaction torques that occur during multijoint movement (Gribble & Ostry, 1999; Sainburg *et al.*, 1993; Sainburg *et al.*, 1995) was responsible for negating the effect these adaptations should have had on the kinematics of the hand.

While the current results provide support for the hypothesis of a proprioceptive deficiency in MDI, it is important to note that there are several alternative explanations that could also account for the data. One possibility is that the dysfunctional passive mechanics of the glenohumeral joint inhibit or interfere with the appropriate implementation of the motor commands sent by the CNS, such that there is a continual mismatch between the efference copy and the sensory reafference. It is also reasonable to infer that pathological joint mechanics would necessarily require atypical patterns of shoulder muscle activation, and a recent study has suggested that this is the case for the rotator cuff and several other shoulder girdle muscles (Barden *et al.*, 2005a). However, this explanation is not inconsistent with deficient proprioception, in that pathological joint mechanics could give rise to decreased peripheral afferent stimulation. Another possibility is that the presence of joint instability produces an excessive amount of background neural "noise", which decreases the signal to noise ratio of the reafference, thereby limiting the fidelity of the internal models (Harris & Wolpert, 1998). Finally, it is conceivable that the pain and apprehension associated with glenohumeral instability creates long-term adaptations in the internal models, such that the observed motor behaviour reflects the development of a protective mechanism. In the context of motor prediction and internal models, it is interesting to consider what responses might occur if the CNS were to predict pain as one of the consequences of performing a motor act.

In conclusion, this study found that the motor behaviour of subjects with MDI, as determined by the hand trajectory inter-repetition errors and velocity profile kinematics of repetitive reaching movements to remembered proprioceptive targets, was significantly different from control subjects with normal, healthy shoulders. The results are consistent with the hypothesis that subjects with MDI possess a proprioceptive deficiency in the symptomatic shoulder, which compromises the neuromuscular control of the upper limb. Future studies should continue to investigate proprioceptive reaching task performance in subjects with MDI using a combination of electromyographic and inverse dynamics procedures to ascertain the neuromuscular and biomechanical mechanisms responsible for the development of this condition.

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6

The effect of elastic bracing on reaching to proprioceptive targets in subjects with and without shoulder instability

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In preparation for submission to the Archives of Physical Medicine and Rehabilitation

Abstract

The effect of an elastic shoulder brace on the ability of subjects with and without multidirectional shoulder instability (MDI) to use proprioception to perform repetitive goal-directed reaching tasks was investigated. Subjects with and without MDI were blindfolded and instructed to return the hand to a remembered proprioceptive target as accurately as possible. Hand position errors and velocity profile kinematic parameters assessed the performance of three distinct reaching tasks: overhead, scapular plane and humeral external rotation with abduction. Subjects performed ten repetitions for each of two task trials, one with and one without a brace. The results demonstrated that wearing an elastic brace had no effect on the hand positioning accuracy or movement kinematics of subjects in either group. However, hand inter-repetition error decreased for both groups as the tasks were repeated. Cycle related kinematic parameter modifications were also observed for subjects in the non-instability group. The intra-task results suggested that hand positioning improved as a consequence of enhanced motion-dependent proprioception, which also facilitated kinematics modifications in non-instability subjects. The cycle related changes to motor output suggest that elastic bracing has no effect on dynamic upper limb proprioception and provides no immediate functional benefit to reaching task performance.

6.1 Introduction

Elastic braces are used in a variety of clinical and sporting environments and are believed to augment joint stability by enhancing proprioceptive input (Birmingham *et al.*, 2000; Jerosch & Prymka, 1996b; McNair *et al.*, 1996; Perlau *et al.*, 1995; Wu *et al.*, 2001). The tight fitting compressive action of the brace ostensibly facilitates an increased discharge of cutaneous, joint and/or muscle spindle afferents (Perlau *et al.*, 1995; Wu *et al.*, 2001). Collectively, these peripheral afferents provide proprioceptive input, which is essential to maintaining joint stability (Jerosch & Prymka, 1996a; Lephart *et al.*, 1997) and controlling multijoint limb movement (Bevan *et al.*, 1994; Cordo *et al.*, 1994; Ghez *et al.*, 1990; Ghez & Sainburg, 1995; Hasan, 1992; Sainburg *et al.*, 1993; Sainburg *et al.*, 1995; Smith & Zernicke, 1987). Numerous studies indicate that deficient proprioception is related to joint instability at the ankle (Richie Jr., 2001), knee (Beynnon *et al.*, 1999; Iwasa *et al.*, 2000) and shoulder (Barden *et al.*, 2004; Blasier *et al.*, 1994; Lephart *et al.*, 1994; Warner *et al.*, 1996). Consequently, if elastic braces have the capability to immediately enhance joint proprioception, then their long-term use has the potential to provide considerable therapeutic benefit to individuals with joint instability related conditions.

Studies of the effects of elastic bracing on proprioception at the ankle and knee have relied primarily on passive joint repositioning and motion detection tests to assess proprioceptive function. For the knee joint, subjects with healthy knees (Birmingham *et al.*, 1998; McNair *et al.*, 1996; Perlau *et al.*, 1995), with ACL injury (Jerosch & Prymka, 1996b) and with ACL reconstructed knees (Lephart *et al.*, 1992; Wu *et al.*, 2001)

demonstrated greater sensitivity on motion detection tests, and greater accuracy on passive positioning tests, when wearing an elastic brace. Conversely, several other studies have shown that wearing an elastic brace has no effect on knee joint proprioceptive acuity or stability (Beynnon *et al.*, 1999; Beynnon *et al.*, 2002; Risberg *et al.*, 1999). While conflicting evidence exists with respect to the effects of elastic bracing on knee joint proprioception, to our knowledge, this relationship has not been studied as it applies to the glenohumeral joint and to shoulder instability.

Glenohumeral stability is achieved via the combined contribution of passive joint structures and dynamic neuromuscular control (Nyland *et al.*, 1998). Multidirectional instability (MDI) is a particularly problematic and complex form of shoulder instability that involves subluxation and/or dislocation of the glenohumeral joint in more than one direction (Beasley *et al.*, 2000). The pathology of MDI includes a loose and redundant joint capsule that causes recurrent pain and instability, particularly in the midrange positions of glenohumeral motion (An, 2000; Arendt, 1988; Beasley *et al.*, 2000; Mallon & Speer, 1995; Schenk & Brems, 1998). Midrange stability is highly dependent on rotator cuff activation, which in turn is mediated directly or indirectly by proprioceptive input (Blasier *et al.*, 1994; Brindle *et al.*, 1999; Lephart *et al.*, 1994). Consequently, maladaptive rotator cuff activation as a result of deficient proprioception has been proposed as one of several contributing factors to the etiology and/or development of MDI (Barden *et al.*, 2004; Blasier *et al.*, 1994; Lephart *et al.*, 1994; Warner *et al.*, 1996).

Given the potential of elastic bracing to improve joint stability via enhanced proprioception, it is surprising that the effects of elastic bracing on joint proprioceptive acuity at the shoulder have not been investigated, either in subjects with healthy

shoulders or in subjects with shoulder instability. The purpose of this study, therefore, was to investigate the immediate effect of an elastic shoulder brace on the ability of subjects with normal, healthy shoulders and with multidirectional shoulder instability, to use proprioception to perform a series of functional, goal-directed reaching tasks. It was hypothesized that the brace would enhance dynamic upper limb proprioception for subjects in both groups, such that modifications to the movement kinematics of the hand would result in improved hand trajectory endpoint positioning errors.

6.2 Materials and Methods

Subjects

Seven subjects with MDI (5 male, 2 female; age = 27.0 ± 8.9 yrs.) and 8 subjects with normal, healthy shoulders (5 male, 3 female; age = 23.4 ± 3.3 yrs.) participated in the study. Ten of the 14 shoulders in the instability group were symptomatic for MDI (3 bilateral and 4 unilaterally symptomatic subjects) and included an equal number (5) of dominant and non-dominant limbs. The 4 limbs without shoulder instability were not tested in the current experiment. For subjects in the healthy shoulder (asymptomatic) group, both limbs were tested (n = 16). The diagnosis of MDI, and the inclusion of subjects into the instability group, was based on the following criteria: (1) a functionally significant inability to keep the humeral head centered in the glenoid fossa, especially in positions not at the extremes of motion; (2) the absence of an injury mechanism likely to tear the glenohumeral ligaments; (3) spontaneous reductions of translations; (4) glenohumeral translations that duplicated the symptoms of concern to the patient; (5) a

diminished resistance to translation in multiple directions as compared with a normal glenohumeral joint; and (6) an absence of traumatic lesions (Matsen III, 1994). Instability group subjects received no treatment or intervention prior to testing and were not afflicted with any psychiatric or emotional difficulties related to voluntary instability. In addition, they did not possess any confounding genetic or neurological health conditions (for example, Ehlers-Danlos syndrome or syringomyelia). Asymptomatic group subjects had no history of shoulder injuries, complaints or surgery. Subjects in both groups indicated limb dominance and provided informed consent prior to testing. The procedures for the current study were approved by the author's institutional Human Research Ethics Review Board.

Shoulder Brace

Each subject wore an individual, custom-fit elastic shoulder brace made of an extensible material (Dacron and Lycra). The brace was developed by one of the principal investigators, an orthopaedic surgeon specializing in shoulder injuries, for clinical use in the treatment of MDI. Preliminary clinical trial data have shown that patients who wear the brace experience greater recovery rates and reduction of symptoms than patients who do not (R. Balyk, unpublished data, 2001). The standard brace resembles a single sleeved T-shirt, with the sleeve encompassing the upper arm, shoulder and scapula of the symptomatic limb (Figure 6.1). Prior to the fabrication of the brace, arm and trunk anthropometric measurements were obtained from each subject so that the brace would fit tightly and comfortably around the upper arm and chest. Subjects with bilateral instability received a modified, double-sleeved brace that was made to fit over both shoulders.

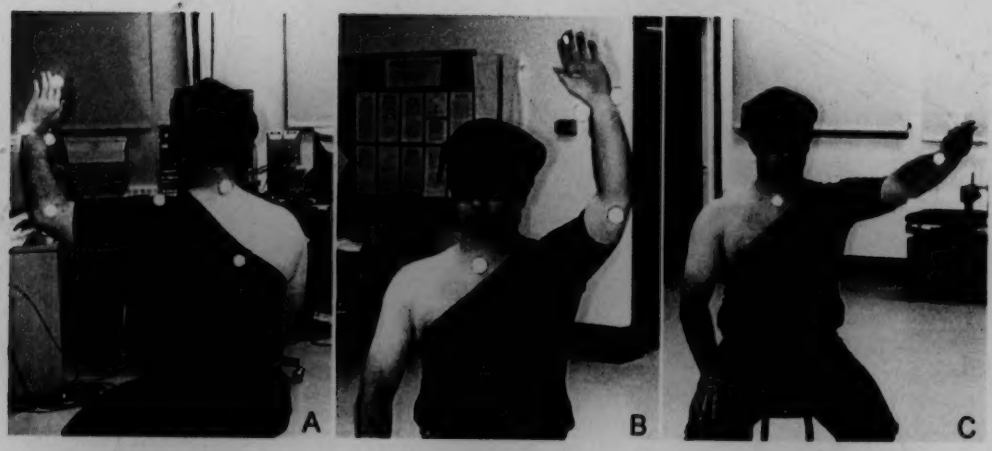

Figure 6.1 A-C: *A shoulder instability subject demonstrates the endpoint position limb configuration for each task: A) shoulder external rotation and abduction, B) overhead reaching and C) scapular plane pointing. The subject is shown wearing the elastic brace, which resembles a single-sleeved T-shirt. The white spheres are the retroreflective markers tracked by the motion analysis system.*

Apparatus

Subjects were blindfolded and seated in the centre of a rectangular test area bounded by four infrared video cameras arranged in an X-configuration (one in each corner). Nine retroreflective markers (1 cm in diameter) were secured to the following landmarks on each subject: the suprasternal notch, the ends of the spinous processes of C5 and T7, the lateral aspect of the acromion, the medial and lateral epicondyles of the humerus, the radial and ulnar styloid processes, and the head of the metacarpophalangeal joint of the middle finger (Figure 6.1). These markers permitted the motion of the upper limb to be captured and recorded without interruption using a three-dimensional optical capture

motion analysis system (Vicon Motion Systems, Lake Forest, CA). In the current study, only hand marker coordinate positions were used in the analysis. Subjects were tested with a complete upper limb marker set so that more extensive analyses could be conducted at a later date.

Rationale

The experimental protocol involved performing unconstrained, repetitive, three-dimensional reaching movements to an actively defined remembered proprioceptive target. The term proprioception refers to all of the peripheral afferent input used by the central nervous system (CNS) to control movement and generate the sensation of joint motion and position (Gandevia, 1996). This study used the motion and position of the arm endpoint (the hand) to quantify reaching task performance. Numerous studies indicate that the CNS uses a hand-centred coordinate system to plan and execute reaching movements (Atkeson & Hollerbach, 1985; Flash & Hogan, 1985; Kaminski & Gentile, 1989; Wolpert *et al.*, 1995). As a result, arm endpoint position errors (referring to the hand as the endpoint of the effector) are typically used to evaluate the motor performance of goal-directed reaching tasks (Adamovich *et al.*, 1998; Darling & Miller, 1993; Flanders & Soechting, 1990; Soechting & Flanders, 1989). Analysis of the hand provides a simple and effective way to determine the degree to which the task objective was obtained, and to assess the means by which this end was achieved (the manner in which the CNS controls the motion of the hand through the rotation of the shoulder and elbow joints).

This study employed an active, rather than a passive, positioning test because the specific conditions that characterize MDI, midrange instability during active muscle contraction, are most effectively assessed with functional measures that quantify the ability to use proprioception to control the movement of the upper limb. Passive proprioceptive acuity has been shown to be least sensitive in the midrange positions of glenohumeral motion (Blasier, *et al.*, 1994; Janwantanakul *et al.*, 2001), thus making the sensation of proprioception evoked by a passive test potentially less relevant to the specific circumstances in which MDI occurs.

Protocol

Subjects performed three separate reaching tasks without vision to targets in different locations of the physical workspace. The tasks were: (1) overhead reaching, (2) scapular plane pointing and (3) shoulder external rotation with abduction (Figure 6.1). Overhead reaching required the arm to be raised to an upright position above the head, with the arm perpendicular to the horizontal plane. Scapular plane pointing consisted of elevating the arm in the scapular plane (approximately 30° anterior to the coronal plane) with the arm parallel to the horizontal. Shoulder abduction with external rotation involved simultaneous external rotation and abduction of the shoulder to a position approaching full external rotation and 90° of shoulder abduction (a motion similar to performing a right-hand turn signal). This movement was performed primarily at the shoulder (the elbow was maintained in approximately 90° of flexion) and was selected because it involves motion about the longitudinal axis of the humerus (internal/external rotation) and because it represents a potentially challenging position for subjects with MDI (the

clinical apprehension test position). Overhead reaching and scapular plane pointing were selected because of their familiarity (subjects were not required to learn any novel tasks or complex motor skills) and generalizability to the wide repertoire of multijoint reaching movements routinely performed in everyday life (reaching to grasp objects above and at various positions in front of and lateral to the body).

Methods

To assess the immediate effect of the brace on proprioceptive reaching task performance, subjects in both groups performed two randomized series of trials, one series with the brace and one without. Asymptomatic group subjects, as well as subjects with bilateral MDI, completed a total of 12 trials (2 braced conditions x 3 tasks x 2 limbs), which consisted of 120 repetitions per subject (12 trials x 10 repetitions per trial). Subjects with unilateral MDI used the symptomatic limb to complete a total of 6 trials (2 braced conditions x 3 tasks) to equal 60 movement repetitions per subject (6 trials x 10 repetitions per trial). All the trials for one test condition (braced or non-braced) were completed before beginning the next series of trials. Movement demonstrations and verbal instructions concerning the specifics of the test protocol were given to each subject at the beginning of the session. Subjects were instructed to move the arm in a specified manner and to self-select a movement speed that would enable them to reproduce the target position with the greatest accuracy. Subjects were also instructed to maintain a constant angle at the wrist joint and to perform a minimum of three practice repetitions until they were comfortable with each task. The practice repetitions were performed at the beginning of each trial once the subjects had been blindfolded. Immediately following

the practice repetitions, the subjects were instructed to move the limb to the appropriate workspace location for that trial (overhead, scapular plane or external rotation/abduction) and to select a virtual target position with the hand (virtual, because no actual physical target was used). The objective of the task was to return the hand to the actively defined proprioceptive target as accurately as possible using the movement specified for that trial. The limb was maintained in the target position for approximately 2 to 3 seconds until the subject was given the command to start the trial. The arm was lowered to a position along the midline of the ipsilateral thigh just below the hip before being returned to the target position. No augmented feedback of any kind (verbal encouragement or knowledge of results) was given concerning the accuracy of each movement repetition or trial. Subjects were instructed not to count, to complete all ten repetitions of the trial without interruption and were told by the experimenter when to stop.

Data Analysis

The three-dimensional hand marker position data were smoothed with a zero-lag, 2nd order Butterworth low-pass digital filter (The Mathworks Inc., Natick, MA) with a cutoff frequency of 6 Hz (Winter, 1990). This procedure smoothed the data so that any subjectivity in determining the endpoint of the trajectory for each repetition was eliminated. These endpoints (approximations to the target) were determined by graphically viewing a three-dimensional representation of the hand marker trajectory to obtain a visual estimate of the direction reversal point. The hand marker coordinates for the appropriate video field that corresponded to the identified endpoint were used to calculate the errors for that repetition. To corroborate the coordinates identified by the

trajectory reversal method, endpoint positions using the minima of the three-dimensional tangential hand velocity profiles were also obtained. There were no discrepancies between the two methods.

To quantify the motor performance outcome of each task, two aggregate (*x*, *y* and *z* directions combined) three-dimensional error measures were calculated. The target position error (TPE) quantified the difference between each endpoint position and the virtual target, thus providing a relative measure of positioning accuracy with respect to the task objective. The inter-repetition error (IRE) quantified the difference between each endpoint position and the previous endpoint position, which provided a relative measure of positioning accuracy with respect to the most recent or updated set of movement conditions. The formulae for calculating mean TPE and IRE for each trial were:

$$\text{Mean TPE} = \left(\sum_{n=1}^{10} \sqrt{(X_n - X_o)^2 + (Y_n - Y_o)^2 + (Z_n - Z_o)^2} \right) \div 10$$

$$\text{Mean IRE} = \left(\sum_{n=1}^{10} \sqrt{(X_n - X_{n-1})^2 + (Y_n - Y_{n-1})^2 + (Z_n - Z_{n-1})^2} \right) \div 10$$

where *X*, *Y* and *Z* represent global coordinate positions in 3D orthogonal space, o = the target position coordinate and n = the repetition number. Both error calculation methods allow the error term for each direction to contribute equally to the total, so that the location of the point in space with respect to the origin of the axis did not bias the result. The magnitude of the resultant hand marker velocity vector, with components along all three orthogonal axes, was defined as the tangential velocity (Adamovich *et al.*, 1998; Milner & Ijaz, 1990). The hand velocity profile is the time history of the tangential

velocity for the reaching phase of each movement repetition (Figure 6.2). To quantitatively compare hand velocity profile kinematics between the braced and non-braced conditions in each group, four kinematic parameters based on the profiles were calculated: peak velocity (PV), average acceleration (AA), average deceleration (AD) and the acceleration time (AT) ratio. The acceleration time ratio (ATR) was calculated as a proportion of the movement time (MT) and therefore quantified the relative proportion of the acceleration phase of the movement (ATR = AT/MT).

For each group, a three-way (brace condition x task x dominance) analysis of variance (ANOVA) with repeated measures (repetition) was conducted for each dependent variable: TPE, IRE, PV, AA, AD and ATR. Statistical significance was set at an α level of 0.05.

6.3 Results

For both subject groups, the results indicated that the immediate application of an elastic shoulder brace had no effect on the endpoint positioning accuracy or movement kinematics of reaching to remembered proprioceptive targets. The group mean values for the braced and non-braced conditions for each task are presented in Table 6.1. There were no significant main effects for bracing for any of the position error or kinematic parameters (Table 6.2). In addition, there were no significant interaction effects between bracing and task or bracing and dominance for any of the parameters in either group (Table 6.2). The results demonstrated that the use of an elastic shoulder brace was of no consequence in terms of achieving the task objective, which was to use proprioception

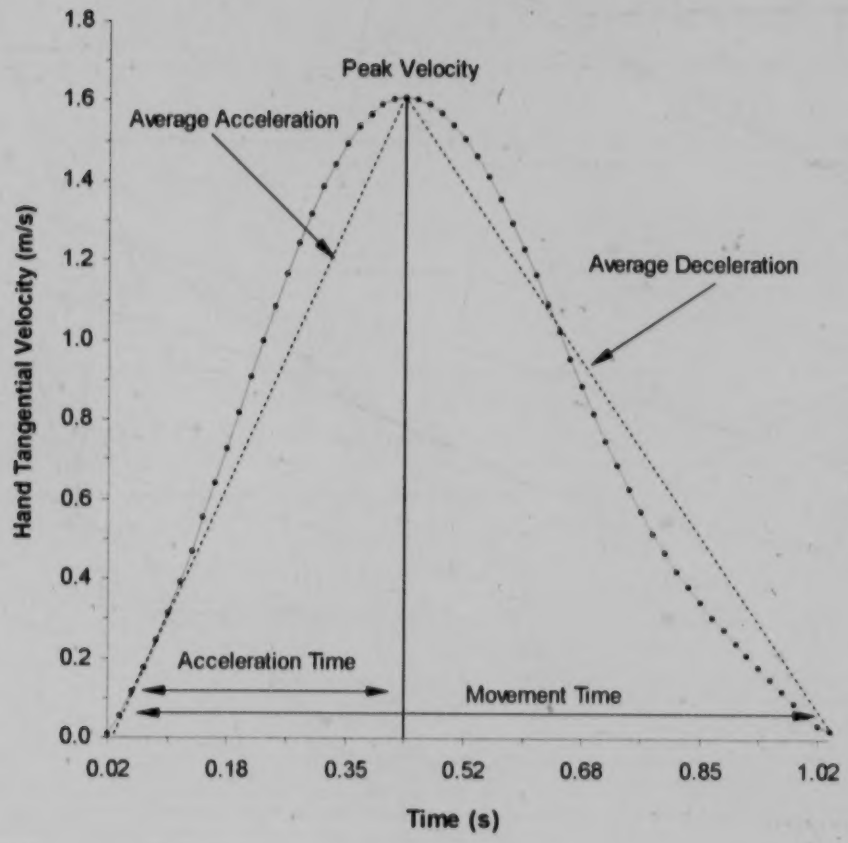

Figure 6.2: *Representative three-dimensional tangential hand velocity profile and associated kinematic parameters.*

to accurately reproduce the remembered proprioceptive target for each task. The results also showed that the brace was of no consequence in terms of achieving a consistent endpoint position (Figures 6.3 and 6.4), which was determined by the endpoint position error relative to the previous repetition (IRE). In contrast to the effects of bracing, the

Table 6.1: Asymptomatic and Instability Group Mean (95% Confidence Interval) Endpoint Position Errors and Kinematic Parameters

	Asymptomatic (n = 7)		Instability (n = 8)	
	Non-braced	Braced	Non-braced	Braced
Target Position Error (mm)				
External Rotation/Abduction	71.26 (5.92)	76.99 (5.26)	80.26 (7.79)	74.07 (5.86)
Overhead	78.71 (5.99)	81.28 (5.71)	65.80 (4.90)	68.15 (5.53)
Scapular Plane	72.44 (5.97)	69.79 (5.33)	69.68 (5.26)	68.10 (6.78)
Inter-Repetition Error (mm)				
External Rotation/Abduction	38.44 (3.83)	39.24 (3.40)	38.69 (4.76)	37.23 (4.46)
Overhead	37.73 (3.41)	36.52 (3.28)	35.95 (4.24)	39.86 (4.49)
Scapular Plane	34.88 (3.16)	30.69 (2.81)	37.09 (4.34)	41.25 (4.29)
Peak Velocity (m/s)				
External Rotation/Abduction	1.62 (0.04)	1.62 (0.04)	1.56 (0.05)	1.51 (0.05)
Overhead	1.77 (0.05)	1.73 (0.05)	1.59 (0.05)	1.66 (0.06)
Scapular Plane	1.42 (0.04)	1.32 (0.04)	1.37 (0.03)	1.35 (0.04)
Average Acceleration (m/s^2)				
External Rotation/Abduction	3.10 (0.15)	3.13 (0.14)	3.28 (0.11)	3.07 (0.10)
Overhead	3.68 (0.14)	3.41 (0.13)	3.24 (0.13)	3.38 (0.14)
Scapular Plane	2.97 (0.12)	2.71 (0.13)	3.03 (0.11)	2.77 (0.13)
Average Deceleration (m/s^2)				
External Rotation/Abduction	-2.04 (0.15)	-1.97 (0.14)	-2.00 (0.11)	-1.95 (0.10)
Overhead	-2.00 (0.14)	-1.95 (0.13)	-1.87 (0.13)	-1.91 (0.14)
Scapular Plane	-1.87 (0.12)	-1.55 (0.13)	-1.91 (0.11)	-1.92 (0.13)
Acceleration Time Ratio (AT/MT)				
External Rotation/Abduction	0.396 (0.008)	0.384 (0.008)	0.386 (0.008)	0.395 (0.008)
Overhead	0.345 (0.007)	0.360 (0.007)	0.363 (0.007)	0.363 (0.008)
Scapular Plane	0.380 (0.007)	0.353 (0.007)	0.387 (0.008)	0.407 (0.009)

repetitive nature of the task was associated with significant position error changes for subjects in both groups. Specifically, significant within subject main effects were found

Table 6.2: Repeated Measures Analysis of Variance Results for Asymptomatic and Instability Group Endpoint Position Errors and Kinematic Parameters

	Asymptomatic		Instability	
	F value	*p* value	F value	*p* value
Target Position Error (mm)				
Brace	0.117	0.734	0.039	0.845
Brace * Dominance	0.214	0.645	0.355	0.554
Brace * Task	0.154	0.858	0.108	0.898
Inter-Repetition Error (mm)				
Brace	0.550	0.460	0.495	0.485
Brace * Dominance	2.114	0.150	0.120	0.730
Brace * Task	0.801	0.452	0.451	0.640
Peak Velocity (m/s)				
Brace	0.506	0.479	0.000	0.999
Brace * Dominance	0.066	0.798	0.095	0.760
Brace * Task	0.300	0.742	0.264	0.769
Average Acceleration (m/s^2)				
Brace	0.427	0.515	0.114	0.737
Brace * Dominance	0.274	0.602	0.524	0.473
Brace * Task	0.164	0.849	0.219	0.804
Average Deceleration (m/s^2)				
Brace	0.645	0.424	0.001	0.974
Brace * Dominance	0.043	0.836	0.168	0.684
Brace * Task	0.211	0.811	0.030	0.971
Acceleration Time Ratio (AT/MT)				
Brace	0.117	0.734	1.196	0.280
Brace * Dominance	0.214	0.645	0.295	0.590
Brace * Task	0.154	0.858	0.582	0.563

for repetition number for both TPE [Instability group: F (9, 405) = 4.81, $p < 0.001$; Asymptomatic group: F (9, 729) = 10.63, $p < 0.001$] and IRE [Instability group: F (9, 405) = 14.02, $p < 0.001$; Asymptomatic group: F (9, 729) = 28.92, $p < 0.001$]. Target

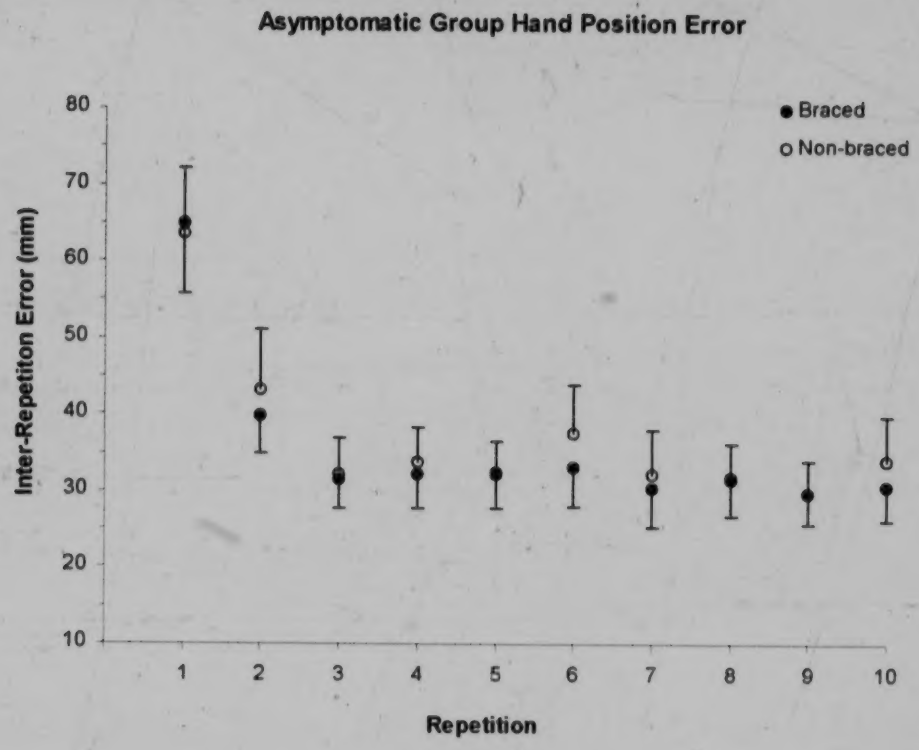

Figure 6.3: *Braced and non-braced mean endpoint inter-repetition errors (pooled across the three tasks) for asymptomatic group subjects. The effect of bracing on the endpoint inter-repetition error was not significant ($p > 0.460$). Repeating the movement resulted in decreased inter-repetition error for both the braced and non-braced conditions ($p < 0.001$). Error bars = 95% confidence interval (CI).*

position errors increased while inter-repetition errors (Figures 6.3 and 6.4) decreased significantly as successive repetitions of the task were performed. These repetition effects

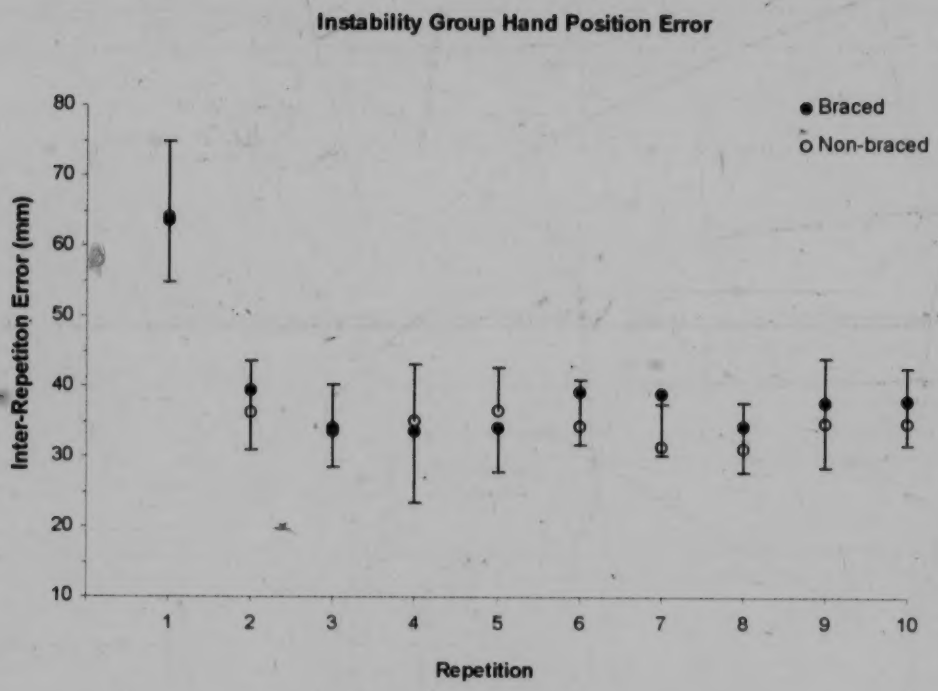

Figure 6.4: *Braced and non-braced mean endpoint inter-repetition errors (error bars = 95% CI) for instability group subjects. Like the asymptomatic group, bracing had no affect on the hand inter-repetition errors of subjects with MDI ($p > 0.485$). Similarly, repeating the movement significantly decreased inter-repetition error for both the braced and non-braced conditions ($p < 0.001$).*

were independent of any other effects, in that no significant interactions were found between repetition and bracing, repetition and task or repetition and dominance in either group. In addition to the non-significant findings for position error, the results also demonstrated that wearing a brace had no effect on the motion of the hand as it was returned to the target. The reaching phase kinematics, as described by the peak velocity,

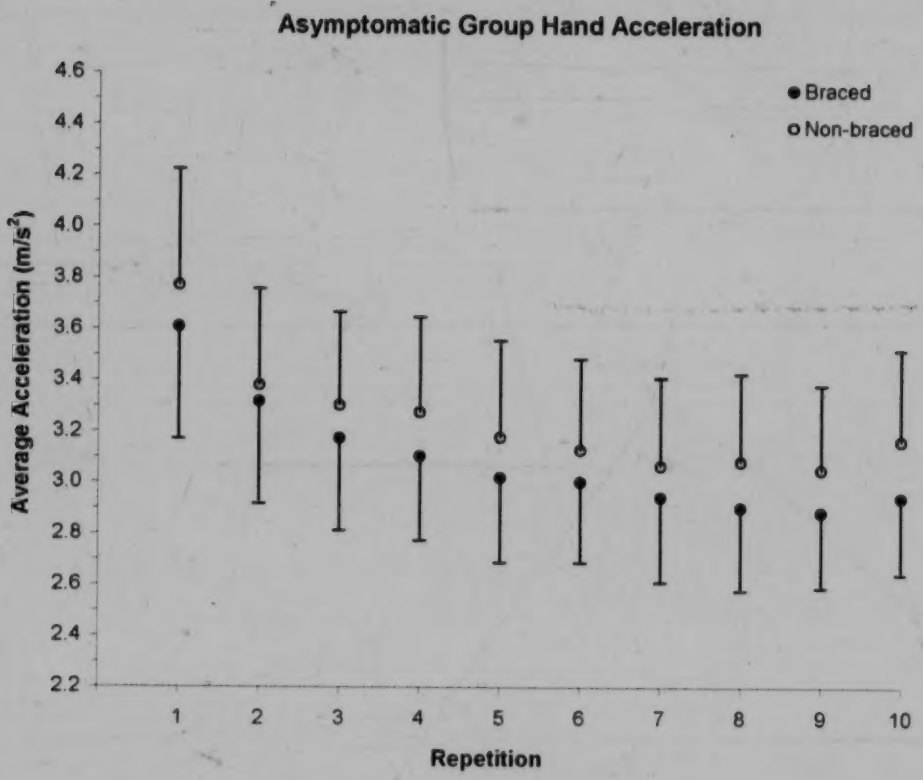

Figure 6.5: *Braced and non-braced hand acceleration values for each repetition (error bars = 95% CI) for asymptomatic group subjects. Wearing the brace had no effect on the average acceleration of any of the repetitions ($p > 0.515$). Both braced and non-braced average acceleration decreased significantly as the trial progressed ($p < 0.001$).*

average acceleration, average deceleration and acceleration time ratio parameters, were the same for the braced condition as the non-braced condition for subjects in both groups (Table 6.2). However, the kinematic parameters changed significantly as the task was repeated for subjects in the asymptomatic group, but in contrast to the position error results, did not change for subjects in the instability group. As the number of movement

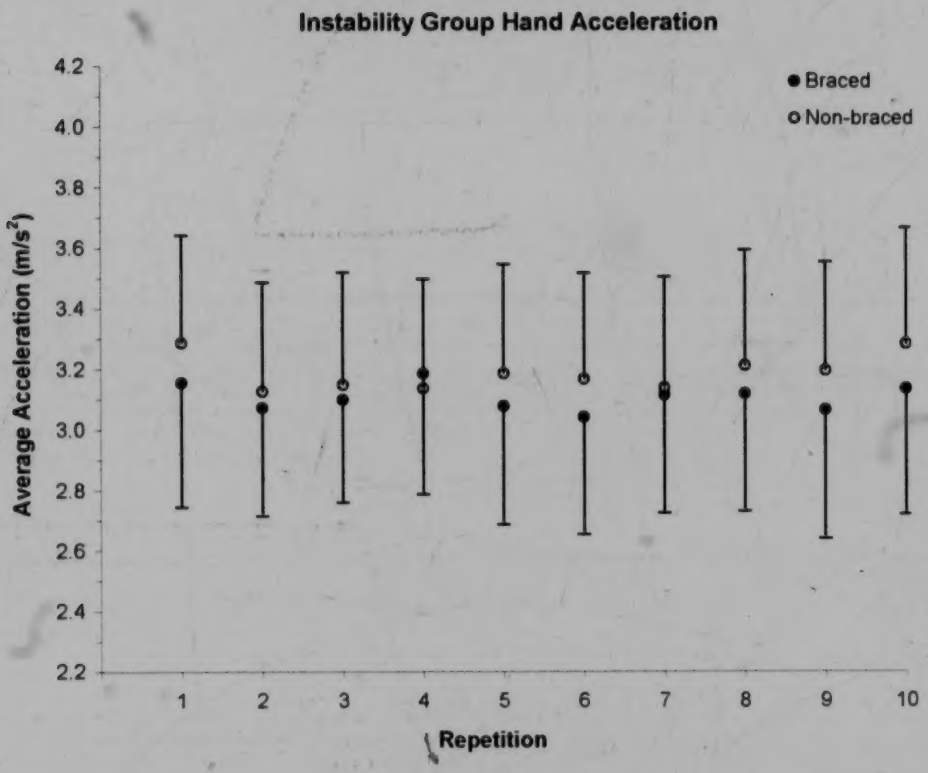

Figure 6.6: *Braced and non-braced hand acceleration values for each repetition (error bars = 95% CI) for instability group subjects. The average acceleration for the braced condition was not significantly different from the non-braced condition (p = 0.737). Unlike subjects in the asymptomatic group, the average acceleration values (braced and non-braced) for subjects with MDI did not change as the movement was repeated (p = 0.481).*

repetitions increased, asymptomatic group subjects demonstrated a significant decrease in PV [F (9, 738) = 38.42, $p < 0.001$], AA [F (9, 738) = 34.30, $p < 0.001$], AD [F (9, 738) = 9.960, $p < 0.001$] and a significant increase in ATR [F (9, 738) = 10.63, $p < 0.001$],

whereas subjects in the instability group did not [PV: F (9, 405) = 1.65, $p > 0.14$; AA: F (9, 405) = 0.93, $p > 0.46$; AD: F (9, 405) = 2.08, $p > 0.16$; ATR: F (9, 405) = 2.17, $p > 0.15$]. For the purpose of the current study, which was to investigate the effects of elastic bracing on proprioceptive reaching task performance, the average acceleration data are presented to illustrate repetition differences between the two groups (Figures 6.5 and 6.6). As with the endpoint position error results, there were no significant kinematic parameter interaction effects between repetition and bracing, repetition and task or repetition and limb dominance for either group.

6.4 Discussion

This study investigated the immediate effects of elastic shoulder bracing on the ability of subjects, with and without shoulder instability, to perform three-dimensional, repetitive reaching tasks to remembered proprioceptive targets. It was found that wearing an elastic brace had no effect on the attainment of the task objective (target position accuracy) for subjects in either group. Similarly, bracing had no effect on the consistency of hand positioning error relative to the previous repetition and no effect on the movement kinematics for subjects in either group. There were no particular combinations of variables (for example, a specific task or limb) for which significant differences between braced and non-braced conditions were observed. Consequently, the hypothesis that the brace would enhance dynamic upper limb proprioception, such that modifications to the movement kinematics would result in improved hand positioning error for subjects in both groups, was not supported.

Given the non-significant results of this study, post-hoc power analyses were conducted for each dependent variable in each group. The probability of detecting a significant difference ranged from 10.9 % to 83.3%, and was lowest for the position error parameters in both groups (the mean power for TPE and IRE combined for both groups was 16.2%, with a standard deviation of 4.3%). The effect sizes for these parameters were small because the standard deviations were considerable, even though the combined number of observations for the two conditions (braced and non-braced) was greater than 500 in each group (Table 6.1). For the inter-repetition error, the possibility that the non-significant braced and non-braced differences were the result of a Type II error is unlikely, because the same methodology has been used to identify significant inter-repetition error differences between similar numbers of MDI and control group subjects (Barden *et al.*, 2004). For the kinematic parameters, the probability of detecting a significant difference ranged from 24.4% to 83.3% (the average statistical power for the four parameters in both groups was 55.4%, with a standard deviation of 22.2%). Like the inter-repetition error, the identical methodology (with similar kinematic parameters and the same numbers of subjects) has been used to successfully identify significant differences in movement kinematics between control subjects and subjects with MDI (Barden *et al*, 2005). Consequently, it is most likely that the non-significant differences between the braced and non-braced conditions occurred because the brace had no effect on the motor performance of the tasks, rather than because the analyses lacked sufficient statistical power to detect the presence of significant differences.

The results of the current study are limited in that they apply to the effects of elastic bracing on the use of motion-dependent proprioception to perform multijoint reaching

movements under a specific set of conditions. They do not apply directly or exclusively to joint specific proprioception or to stability at the glenohumeral joint, even though shoulder stability is a necessary prerequisite and ongoing requirement for the neuromuscular control of arm movements. The present results apply to low speed (a consequence of the accuracy constraint) voluntary reaching movements, of the type that are regularly experienced during everyday life (reaching to interact with objects in different spatial locations). The potential effects of elastic bracing on the neuromuscular control and joint stability associated with involuntary reflex actions or of fast, ballistic arm movements are beyond the scope of the current investigation. This study used an active positioning test with realistic, three-dimensional arm movements so that the effects of elastic bracing would have the greatest generalizability to the actual conditions in which proprioceptive input mediates the neuromuscular control of the upper limb. The generalizability of passive test procedures to the conditions in which natural movements occur should be questioned, because active and passive movements have been shown to stimulate synaptic neuronal activity in different parts of the brain (Weiller *et al.*, 1996). Furthermore, even relatively simple movements generate an extraordinary amount of sensory feedback, such that the CNS appears to use a gating strategy to deal with the large influx of reafferent input (Apps, 1999; Chapman, 1994; Collins *et al.*, 1998; Lemon, 1999). Consequently, studies that investigate the effects of orthotic devices (such as elastic and functional braces) on proprioception and neuromuscular control should do so in an appropriate context.

Studies that have used natural locomotor movements to examine the effects of functional knee braces have found that bracing alters lower limb joint kinetics for

walking and running in healthy subjects (DeVita *et al.*, 1996) and in subjects with ACL reconstructed knees (DeVita *et al.*, 1998). Other studies have shown that functional knee braces provide no mechanical benefit to joint stability in the ACL reconstructed (Wu *et al.*, 2001) or ACL deficient knee (Ramsey *et al.*, 2003). Consequently, it is unlikely that an elastic brace would possess sufficient rigidity to mechanically augment joint stability. If orthotic devices such as elastic braces are to benefit joint stability, that benefit must be neurological rather than mechanical, and must confer some advantage to the actual, functional operation of the neuromuscular control system of which joint stability is an integral component. The functional paradigm employed in the current study was developed to assess the effects of elastic bracing on dynamic upper limb proprioception. The test required subjects to reach to proprioceptive targets so the CNS would possess only a proprioceptive (rather than a proprioceptive and a visual) estimate of the initial joint configuration and spatial orientation of the limb. Actively defining the target (without vision) also ensured that the CNS had an initial motion-dependent (dynamic) proprioceptive estimate of the limb dynamics needed to accurately reproduce the movement, as well as a template of the control signals needed to move the limb to the target. The process of actively defining the target required the limb to be maintained briefly (2 to 3 seconds) in the target position immediately prior to movement initiation. This was done so that the CNS obtained a static proprioceptive estimate of the joint forces and torques needed to counteract the gravitational inertial forces that were unique to the specific limb configuration selected by the subject (Adamovich *et al.*, 1999).

Given these conditions, it is important to discuss the non-significant brace results in the context of the task objective, which was to use upper limb proprioception to

repeatedly reproduce an actively defined proprioceptive target. The repetitive nature of the tasks resulted in significant position error differences for all conditions, irrespective of whether a brace was used. The data for subjects in both groups showed that the target position errors increased while the inter-repetition errors decreased as the tasks were repeated. Increased target position error suggests that the original sensorimotor specifications of the target position, which defined the initial limb conditions prior to the execution of the first repetition, became increasingly less relevant as the movement was repeated (Barden *et al.*, 2004). In contrast, the decreased inter-repetition errors, which quantified position error relative to the most recent (or updated) set of movement instructions and conditions (the outcome of the previous movement), suggested that the CNS received updated sensorimotor movement information several times before an optimal level of execution was achieved. Several studies have shown that the modification of motor output by feedforward control mechanisms is based on the presence of updated limb dynamics information provided by motion-dependent proprioceptive input (Ghez *et al.*, 1990; Messier *et al.*, 2003; Sainburg *et al.*, 1995). Consequently, the target position and inter-repetition error results suggest that moving the limb resulted in motion-dependent proprioceptive feedback, which served to update or replace the sensorimotor information obtained during the target selection procedure. The CNS presumably used this updated information to fine-tune the control signals of the movement such that the consistency of hand positioning was affected.

The average acceleration and other kinematic parameter results for subjects in the asymptomatic group supports this interpretation, in that the average acceleration decreased significantly as the tasks were repeated (see Barden *et al.*, 2005 for further

information concerning cycle related changes for hand velocity profile kinematics). The average acceleration parameter characterizes the acceleration phase of the tangential hand velocity profile, which can be used to approximate the level of muscle torque needed to accelerate the arm toward a spatial target (Adamovich *et al.*, 1999). The average acceleration data for subjects in the asymptomatic group could indicate that the CNS overestimated the muscle torque needed to produce the first repetition (possibly because it overestimated the inertial limb forces), and subsequently used motion-dependent proprioceptive feedback to progressively refine this estimate to an appropriate level as the number of movement repetitions increased (Hasan, 1992). In contrast, the average acceleration for subjects in the instability group did not change and was constant throughout the trial. The lack of modifications to this parameter suggests that subjects with MDI may not adjust the level of muscle torque in response to updated limb dynamics information provided by dynamic proprioception (Barden *et al.*, 2005).

Most importantly, these results suggest that repeating the task enhanced dynamic upper limb proprioception, in so far as it was used to control the motion of the hand to accomplish the task objective. The results also demonstrated that the subsequent motor output adaptations were not affected by bracing, which suggests that the brace did not augment proprioceptive input. The discrepancy between the negative findings of the current study and the positive findings of several passive proprioception studies (Birmingham *et al.*, 2000; Jerosch & Prymka, 1996b; McNair *et al.*, 1996; Perlau *et al.*, 1995; Wu *et al.*, 2001) is likely the result of many factors, most notably differences in the experimental procedures and the methods used to quantify proprioception. To our knowledge, the current study is unique because it is the only bracing study to evaluate the

effect of dynamic proprioception on the motor performance of functional, three-dimensional arm movements. These findings are particularly relevant to shoulder instability, because evidence suggests that deficient proprioception is related to the incidence of anterior (Blasier *et al.*, 1994; Lephart *et al.*, 1994; Warner *et al.*, 1996) and multidirectional shoulder instability (Barden *et al.*, 2004). The present results fail to support the idea that an elastic shoulder brace will provide functional neuromuscular benefits (such as improved joint stability) to individuals with MDI through the amelioration of deficient proprioception. Nevertheless, the results do not exclude the possibility that an elastic brace might improve proprioception and stability at the glenohumeral joint in such a way that the motor performance of the arm endpoint was not affected. It is also possible that elastic shoulder bracing might produce functional motor performance benefits under different conditions, namely through long-term use or for different types of movements, such as those involving spinal reflex control.

In conclusion, this study found that elastic shoulder bracing had no immediate effect on the motor performance of arm endpoint positioning, when subjects with and without multidirectional shoulder instability performed repetitive reaching tasks to remembered proprioceptive targets. In contrast, repeating the task produced significant motor performance changes for subjects in both groups, suggesting that initial limb conditions were updated with motion-dependent proprioceptive feedback that was subsequently used to fine-tune the control signals of the movement. Consequently, the effect of the brace on the ability of the CNS to use dynamic upper limb proprioception to perform repetitive goal-directed reaching tasks was either non-existent or inconsequential. Future studies should investigate a variety of test procedures and

movement conditions to determine the immediate and long-term effects of elastic bracing on the neuromuscular control of the upper limb and the joint stability of the shoulder.

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locations specified by Cram and Kasman (1998). The outer edges of the electrodes were placed next to each other to standardize the centre-to-centre inter-electrode distance at 3.2 cm and 3.65 cm. A reference electrode was placed on the olecranon process of the test limb.

Intramuscular electrodes were constructed by inserting single 50 μm, nylon coated, nickel-alloy fine-wires (California Fine Wire Company, Grover Beach, CA) into the cannulas of 1.5 inch, 27-gauge hypodermic needles. The wires were prepared by removing 2 mm of insulation from both ends of the wire. This created a bared recording tip that was bent into a hook after it was inserted into the needle. The needles with the inserted wires were then sent for gas sterilization. To ensure a standardized inter-electrode spacing of 1 cm for the bipolar configuration, a dual-needle insertion technique (one wire per needle) was used to insert the wires into the infraspinatus and supraspinatus muscles in the manner described by Kelly *et al.* (1997). The dual-needle technique has been shown to produce superior signal characteristics and to decrease signal acquisition variability compared with the traditional single-needle technique, in which two fine-wire electrodes are placed into the muscle with a single needle (Kelly *et al.*, 1997). The two electrodes for each muscle were inserted parallel to the muscle fibre direction, 0.5 cm on either side of the identified location points. The identified location point for the supraspinatus was approximately 1.5 cm above the midpoint of the spine of the scapula, and approximately 2.5 cm below the midpoint of the spine of the scapula for the infraspinatus (Kelly *et al.*, 1997). To standardize wire placement depth within the muscle, the needles were inserted until contact was made with either the infraspinous or

supraspinous fossae. Following muscle insertion and the removal of the needles, the wires were securely taped to the skin and crimped onto gold-plated connector pins.

The myoelectric signals were pre-amplified at the skin surface using differential amplifiers with a common mode rejection ratio of 100 dB, an input impedance of 10^9 Ω and a gain of 100. Surface and fine-wire electrodes were connected to an 8-channel connector box on a belt-pack that subjects wore around their waist. The 8 channels (7 EMG channels plus one angular position channel) were connected to a NI DAQCard-1200 A/D card (National Instruments Corporation, Austin, TX). The data were simultaneously sampled at 2500 Hz using LabWindows/CVI 5.0 data acquisition software (National Instruments Corporation, Austin, TX), which stored the binary data on a digital computer for further processing and analysis.

Experimental Procedure

The 7 limbs with symptomatic MDI shoulders were tested along with one randomly selected limb from each subject in the control group. Prior to beginning the test protocol, electrode positions were verified with a series of manual muscle tests designed to isolate each muscle (Kelly *et al.*, 1996). These tests also ensured that the fine-wire electrodes were securely embedded into the infraspinatus and supraspinatus muscles (Morris *et al.*, 1998). The test procedure required subjects to perform the following planar shoulder movements on a KinCom 500H isokinetic dynamometer (Chattex Corporation, Chattanooga, TN): shoulder abduction/adduction, flexion/extension and internal/external rotation. Each subject completed ten consecutive repetitions of each task. Multiple repetitions were conducted to adequately represent the true EMG pattern of each subject

(Yang & Winter, 1984), thereby facilitating the accuracy of the assessment. The ten repetitions of each trial were performed consecutively (i.e., without an extended pause at the movement phase transition), because pilot testing revealed that non-consecutive repetitions did not result in the deactivation of certain muscles once the end of the range of motion had been reached. In some cases this occurred because the limb was maintained in a static position against gravity (shoulder abduction for example), while in other cases it occurred because of voluntary activation on the part of the subject. Consequently, the comparison of electromyograms between groups would have been less accurate with a non-consecutive repetition test. Each subject performed three randomized trials, one for each task (3 trials x 10 repetitions per trial = 30 repetitions per subject). Subjects rested for approximately 5 minutes between trials while the dynamometer configuration was modified. Subjects performed flexion/extension and internal/external rotation from a seated position, and laid on their side to perform shoulder abduction/adduction. Internal/external rotation was tested with the shoulder in 90° of abduction and the elbow flexed at 90°. Each task involved a 90° range of motion: 20°-110° for flexion/extension and internal/external rotation (at 0° the forearm was parallel to the frontal plane) and 10°-100° for abduction/adduction.

The tasks consisted of isotonic, concentric/concentric repetitions with a constant force level of 35 N. The protocol used movements with constant force (isotonic), rather than constant velocity (isokinetic), because constant velocity protocols were found to require excessive force at the beginning of the range of motion, and were not effective in simulating natural arm movements. However, to control movement speed, the angular velocity was limited to 170° per second. The subjects were instructed to reach the speed

limit on each repetition and to maintain this speed throughout the greater part of the range of motion. Numerous practice repetitions were completed to allow the subjects to become familiar with the arm movements and test procedures.

Data Analysis

The data were initially processed using MATLAB 5.3 (The Mathworks Inc., Natick, MA), in which a dual-pass Butterworth digital filter applied band-pass cutoff frequencies of 10-500 Hz and 30-750 Hz for the surface and fine-wire electrodes, respectively. The channel with the angular position data from the KinCom electrogoniometer was low-pass filtered with a cutoff frequency of 6 Hz (Winter, 1990). The raw EMG data (Figure 7.1A) were then processed using an onset detection algorithm created with Visual Basic 6.0 (Microsoft Corp., Redmond, WA). A two second quiet period (5000 samples) prior to the initiation of the trial was used to calculate the mean and standard deviation of the baseline signal. The entire data set for each trial was subsequently full-wave rectified (Figure 7.1B) and transformed into multiples of the baseline standard deviation (SD units). The average standard deviation was then calculated across a 50 millisecond moving average window (125 samples). The window at which the average value exceeded 3 standard deviations was used as the threshold to determine the onset of activation (David *et al.*, 2000; Hodges & Bui, 1996). The actual time of EMG onset corresponded to the point within the identified window at which the signal exceeded and remained above the threshold value. The same process was used to establish the termination of the muscle activity (the time at which the average standard deviation decreased to less than 3 SD units). In all cases, the algorithmic method of determining the

muscle activation and deactivation was verified by a visual inspection of the data. These points in time were then matched to the corresponding angular position values (Figure 7.1A-B) and converted into relative angles associated with the beginning and end of the movement (the contraction was assigned to the movement phase in which the majority of the muscle activity occurred). The onset (ONST) and termination (TERM) angles were designated as positive or negative according to whether they occurred within or outside the designated phase (positive angles occurred within the phase, negative angles occurred without). The duration of the muscle activity as a function of the range of motion (90 degrees for each movement phase) was calculated using the relative onset and termination angles as follows:

$$\text{Range of motion (ROM)} = 90 - (\text{ONST} + \text{TERM})$$

such that if a muscle such as the posterior deltoid (Figure 7.1A) was active primarily during shoulder abduction, with the onset of activity occurring 7 degrees prior to the initiation of abduction (-7) and terminating 8 degrees before the end of abduction (+8), the muscle was active throughout an 89-degree range of motion.

The ROM parameter quantified the duration of the muscle activity (Figure 7.1B), and was not limited to or directly affected by the absolute angular positions that defined the beginning and end of the movement (for example, as shown in Figure 7.1A, abduction commenced with the arm in 10 degrees of abduction and terminated at 100 degrees, and vice versa for adduction). The ONST, TERM and ROM parameters were reported as position-based values (rather than as time-based values) so that they would be independent of movement speed. Consequently, in the case of the example presented in

Figure 7.1A-B, the posterior deltoid (along with the anterior and middle fibres) acted both as an agonist and antagonist to raise and decelerate the arm in abduction and adduction, respectively. This type of dual-phase muscle activation (i.e., occurring across the boundary or direction reversal point of the movement) was possible because the ten-repetitions of each task were performed consecutively.

The primary purpose of the onset detection algorithm was to capture the myoelectric signal and to determine its properties in the time domain. However, it was also possible to quantify the amplitude characteristics of the captured electromyograms by expressing each window value as a percentage of the peak value (Morris *et al.*, 1998; Yang & Winter, 1984). These normalized values were used to calculate the average EMG (the average value of all the activation windows). Consequently, processing the data yielded 4 EMG parameters that were subsequently used for statistical analyses: the normalized average amplitude (AMP), the onset of activation (ONST), the termination of activation (TERM) and the duration of activation in terms of the range of motion (ROM). The first and last repetitions of each trial were omitted from the statistical analyses because the non-continuous nature of these repetitions resulted in muscle activation parameters that were different from the other eight repetitions. For each muscle, a two-way (group x task) analysis of variance (ANOVA) with repeated measures (repetition) was conducted for each dependent variable: AMP, ONST, TERM and ROM. Bonferroni post-hoc tests were used to identify significant main or interaction effects. Probability (p) values less than 0.05 were reported as statistically significant.

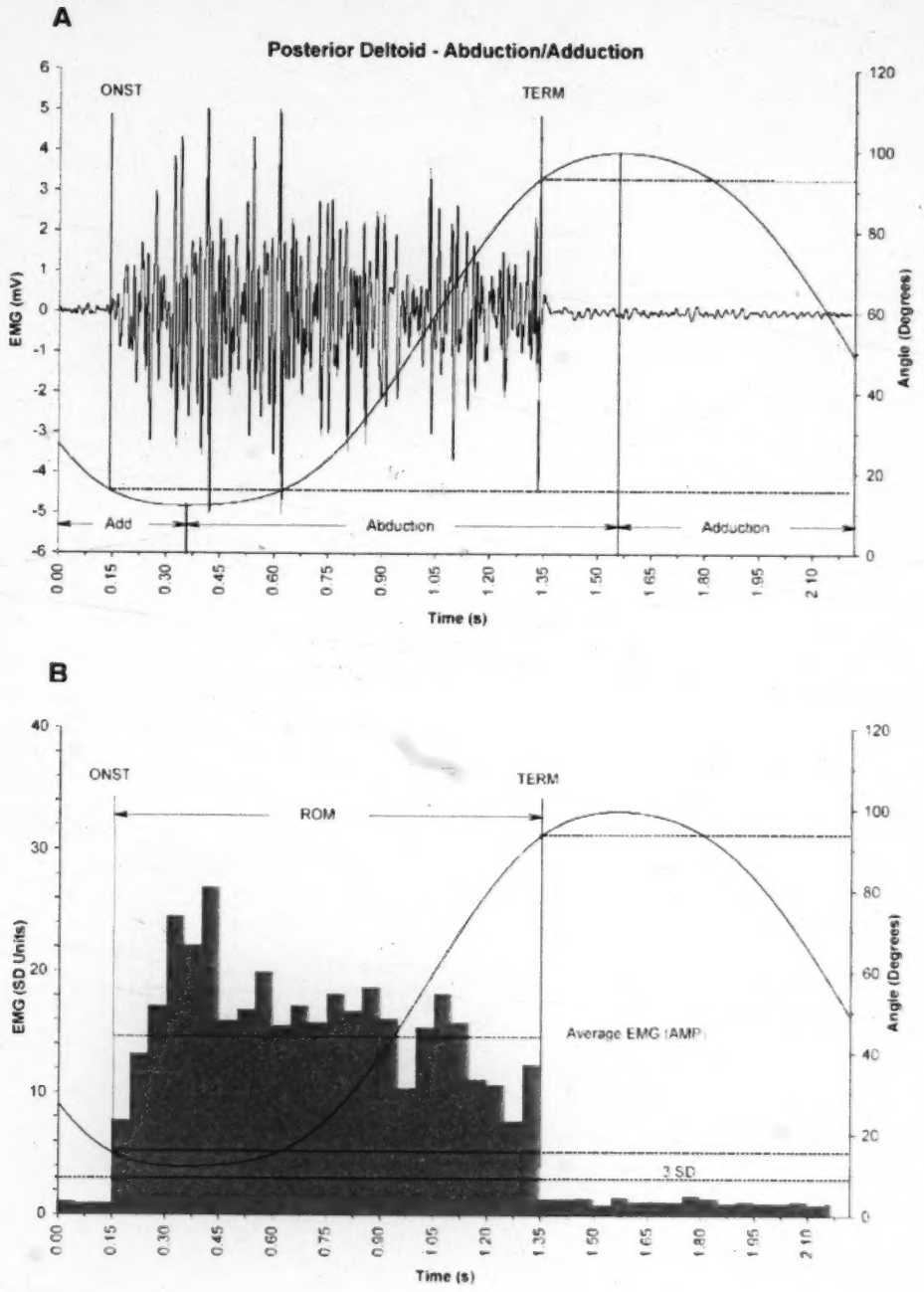

Figure 7.1: *Representative electromyogram for the posterior deltoid during shoulder abduction adduction indicating arm angular position, activation onset (ONST) and*

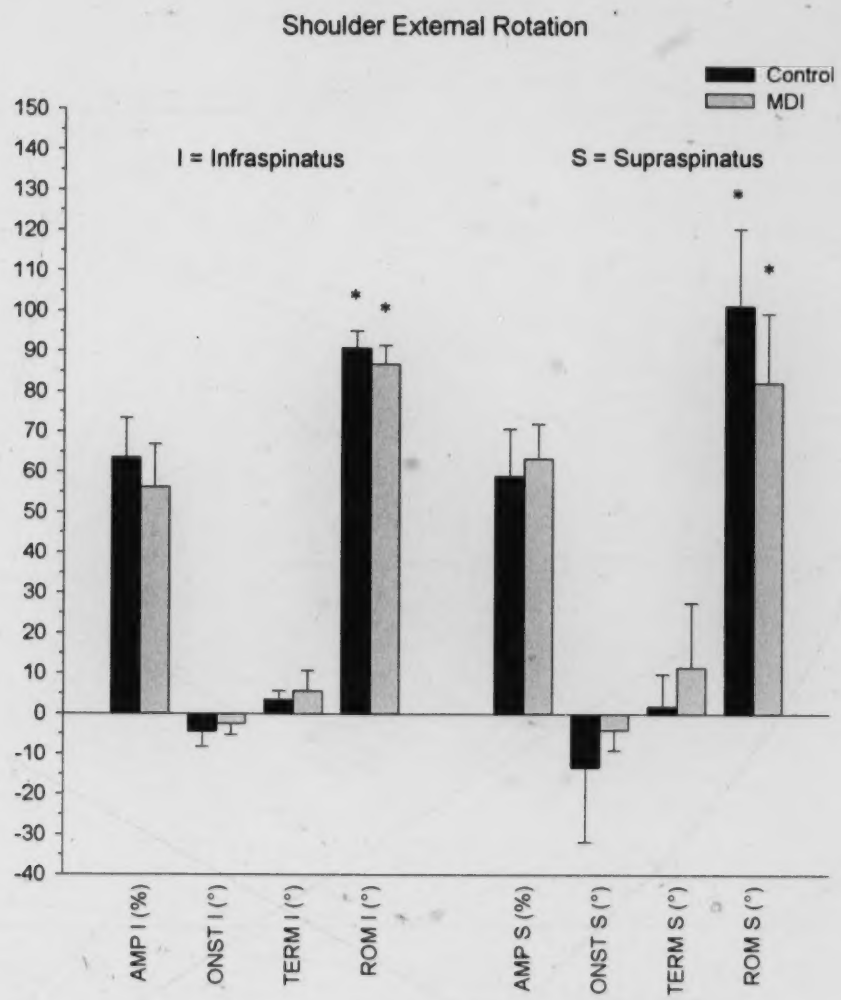

Figure 7.4: *Group mean activation parameter values for the infraspinatus (I) and supraspinatus (S) muscles during shoulder external rotation. The ROM parameter for the infraspinatus (* = $p < 0.01$) and supraspinatus (* = $p < 0.01$) was significantly less for the MDI group. Error bars = standard deviation of the mean.*

indicated by the negative values). The data also demonstrate the basis for the significant interaction, in that for shoulder abduction and flexion the ONST for the MDI group occurred before the ONST for the control group, whereas for external rotation the MDI ONST occurred 34 degrees after the onset for the controls.

For the termination (TERM) parameter, a significant main effect for task was found [F (2, 48) = 5.59, $p < 0.007$], such that the termination of the activity for both groups during external rotation occurred earlier than it did for shoulder abduction ($p < 0.018$) and shoulder flexion ($p < 0.008$). No significant differences between the control group and the MDI group were found for the TERM parameter. For the range of motion (ROM) parameter, the analysis identified a group main effect that approached significance [F (1, 48) = 3.66, $p < 0.06$] as well as a significant group x task interaction [F (2, 48) = 3.83, $p < 0.05$]. Post-hoc testing revealed that there were no significant MDI vs. control differences for any of the tasks, although the difference for external rotation approached significance at $p < 0.09$, such that the subjects with MDI activated the posterior deltoid throughout a smaller range of motion than the controls. The reduced ROM for subjects in the MDI group occurred as a result of the significant delay in the ONST of the activity.

Pectoralis Major

For the pectoralis major, a significant group main effect [F (1, 47) = 6.67, $p < 0.01$] occurred for the AMP parameter. Post-hoc analyses determined that the AMP for the MDI group was significantly less ($p < 0.005$) than the AMP for the control group during shoulder extension (Table 7.2). No significant differences were found for the ONST parameter. Group main effects that approached significance were also found for the

TERM [F (1, 47) = 3.68, $p < 0.06$] and ROM [F (1, 47) = 3.27, $p < 0.07$] parameters. Post-hoc tests revealed that the MDI and control group differences for TERM and ROM were significantly different ($p < 0.01$) when performing shoulder extension, such that the subjects with MDI terminated the activity of the pectoralis major later than the control subjects, so that the pectoralis major was active throughout an extended range of motion (Table 7.2).

7.4 Discussion

The purpose of the current study was to use EMG to investigate the timing and amplitude of shoulder muscle activation in subjects with multidirectional instability (MDI). The hypothesis that the rotator cuff muscles of subjects with MDI would demonstrate temporal differences in activation, compared to control subjects with healthy shoulders, was supported. In the shoulders with MDI, abbreviated activation of the supraspinatus muscle occurred for all tasks in which the muscle was active. Significant activation parameter differences between the control and MDI groups were also found for the infraspinatus, posterior deltoid and pectoralis major muscles. Collectively, the posterior deltoid, infraspinatus and supraspinatus muscles of subjects with MDI demonstrated shortened periods of activation when performing the task of internal/external rotation, despite activation amplitudes that were similar to the controls. In contrast, the activation of the pectoralis major differed from the control group in both the amplitude and time domains when performing shoulder extension. Given the importance of rotator cuff activation to the dynamic stability of the shoulder, the supraspinatus and infraspinatus

muscles will be discussed first, followed by brief discussions of the posterior deltoid and pectoralis major muscles.

Supraspinatus and Infraspinatus

The results demonstrate that for all three test movements, subjects in the MDI group activated the supraspinatus muscle throughout an abbreviated range of motion relative to the controls. The primary cause of the abbreviated activation was premature termination of the muscle activity near the end of the movement in which the muscle was active. The difference in contraction duration was most pronounced for external rotation, in which the timing of the MDI activation was characterized by a delayed onset and a premature deactivation relative to the corresponding positions at which control subjects activated and deactivated the muscle. Studies of rotator cuff activity in subjects with healthy shoulders have shown that the supraspinatus makes an important contribution to external rotation, particularly when the arm is placed in an abducted position. These studies have also shown that the contribution of the supraspinatus increases as the abduction angle at the shoulder is increased (Ihashi *et al.*, 1998; Kronberg *et al.*, 1990).

The results for the infraspinatus demonstrate that the pattern of activation in the MDI group differed from the control group, but only during external rotation. Like the supraspinatus, infraspinatus activity for external rotation in subjects with MDI was confined to a limited range of motion. This limited range occurred because the muscle was activated and deactivated within the normal boundaries set by the control group. At the point of onset, which occurred for both groups just before the end of internal rotation, the activation of the infraspinatus served to decelerate the humerus in preparation for the

transition to external rotation. The data indicate that the infraspinatus was deactivated just before the end of external rotation, when the arm had reached approximately 65° of external rotation relative to the neutral position.

Studies show that the infraspinatus and supraspinatus muscles combine to stabilize the glenohumeral joint when the arm is externally rotated in the abducted position (Itoi *et al.*, 1994). Not surprisingly, this position (which is clinically termed the apprehension test position) is frequently associated with anterior subluxation or dislocation of the glenohumeral joint (Kido *et al.*, 2003). Consequently, it is likely that the premature deactivation of the infraspinatus and supraspinatus muscles in subjects with MDI would seriously compromise the ability of the rotator cuff to provide anterior stability to the glenohumeral joint in the externally rotated and abducted position. The present results suggest that for external rotation, subjects with MDI experience a loss of dynamic stability at a critical point in the range of motion.

The temporal differences in rotator cuff activation contrasted with the non-significant findings in the amplitude domain, which were consistent with the results of Morris *et al.* (2004), who recently conducted the first investigation of shoulder muscle activity in MDI. In comparing the activation levels of the deltoid (anterior, middle and posterior) and rotator cuff muscles (infraspinatus, supraspinatus and subscapularis) in subjects with MDI to control subjects with healthy shoulders, Morris *et al.* (2004) found that there were no significant differences in the amplitude of the peak normalized EMG between the rotator cuff muscles of the two groups. It is interesting that the present EMG amplitude results, along with those of Morris *et al.* (2004), contrast with the study of McMahon *et al.* (1996), who found significantly diminished levels of supraspinatus

activity in subjects with anterior instability during shoulder scaption and abduction movements. Unfortunately, a comparison between the results of this study and the findings of Morris *et al.* (2004) with respect to the timing of the rotator cuff activity is not possible, because Morris *et al.* (2004) did not investigate the temporal characteristics of the recorded electromyograms.

Consequently, the findings of the present study suggest that subjects with multidirectional instability use the same levels of rotator cuff activity as subjects with normal, healthy shoulders, but that the timing of the recruitment is characterized by an asynchronous pattern of activation, especially in relation to the termination of the muscle activity. The current results provide support for the theory that multidirectional instability of the shoulder is associated with a neuromuscular control deficiency that affects the coordinated activation of the rotator cuff (Mallon & Speer, 1995; Schenk & Brems, 1998). The results also suggest that this deficiency is accentuated when the shoulder is externally rotated with the arm in 90 degrees of abduction.

Posterior Deltoid

In the MDI group, the onset of activation for the posterior deltoid was found to be significantly different from the control group during internal/external rotation. Other studies have found significant differences in the levels of muscle activity for the posterior and middle deltoid in subjects with MDI (Morris *et al.*, 2004) and in subjects with anterior instability and generalized joint laxity (Kronberg *et al.*, 1991). In the present study, control subjects activated the posterior deltoid 46 degrees before the end of internal rotation, whereas the onset of activity for the MDI group occurred only 12

degrees before the end of the internal rotation phase was complete. Although the capacity of the deltoid to function as a stabilizer of the glenohumeral joint has not been thoroughly investigated, Kido *et al.* (2003) have shown that all three portions of the deltoid are capable of providing anterior stability to the glenohumeral joint. Consequently, it is likely that the function of the posterior deltoid near the end of internal rotation was to act eccentrically to control the movement and to provide joint stability by reducing or preventing anterior translation of the humeral head. If so, the delay in the onset of the activity in the MDI group would potentially compromise the anterior stability of the glenohumeral joint in this position.

Pectoralis Major

In addition to finding differences in the timing of the activation for the infraspinatus, supraspinatus and posterior deltoid muscles in subjects with MDI, this study also found differences in the activation of the pectoralis major during shoulder extension. The results indicate that, unlike control subjects, the majority of subjects with MDI failed to exhibit a pattern of activation and deactivation for each phase of the task. For flexion/extension, MDI subjects maintained low levels of activity that persisted throughout the greater part of the range of motion of the entire task (180 degrees), whereas control subjects demonstrated distinct patterns of activation and deactivation for each of the separate phases. This particular phenomenon was not observed in the MDI group for either the abduction/adduction or internal/external rotation tasks. The results show that the pattern of activity between the two groups differed because of the termination of the muscle activity, rather than the onset. Subjects with MDI sustained the activity of the pectoralis

results, which showed that the velocity, acceleration and deceleration of the hand decreased after the first repetition.

In comparing the proprioceptive reaching task performance of healthy subjects to the subjects with MDI, it was found that subjects with MDI performed similarly with respect to hand trajectory drift and movement extent but the trajectory endpoint inter-repetition errors were significantly greater than those of the control group. This finding demonstrated that subjects with MDI were unable to achieve the same level of movement reproduction capability as subjects in the control group. It was also demonstrated that the hand velocity profile kinematics of the MDI group remained unchanged as the tasks were repeated, which contrasted with the findings of the control group.

In general, the inferences that can be made about the processes involved in controlling the arm to perform reaching movements under these conditions are consistent with the hypothesis that subjects with MDI have a proprioceptive deficiency that compromises the neuromuscular control of the symptomatic shoulder. More specifically, a potential theory to account for the data is that as the subjects with MDI repeatedly returned the limb to the proximity of the actively defined target position, the ability of peripheral afferent receptors to transform the resulting joint motion into sensory feedback was inadequate, such that the sensory reafference contained insufficient information. Consequently, after comparison with the sensory prediction, the information transmitted to the internal inverse dynamics model (the error signal) could have been of lower quality, such that the model did not receive the appropriate update concerning the recently completed act. This in turn would have an effect on the subsequent motor commands issued by the controller for the next repetition, and therefore on the centrally

derived efference copy as well. In general terms, a deficiency in the peripheral afferent input would have a deleterious affect on the entire sensorimotor control system, including the fidelity of predictions for subsequent movements produced by the internal forward model via the efference copy mechanism. Given such a scenario, which although speculative is consistent with the results, it is reasonable to conclude that a lower level of precision for motor prediction would equate to a lower level of precision for movement reproduction. It is also important to reiterate that the hypothesis of deficient proprioception is consistent with the pathology and symptoms of MDI, namely capsuloligamentous laxity and compromised dynamic stability in the midrange positions of glenohumeral motion. A potential mechanism for deficient proprioceptive feedback would be the lack of capsuloligamentous tension, which results in an inadequate stimulation of local proprioceptive afferents and/or a lack of muscle spindle sensitivity (fusimotor set) via feedforward alpha-gamma coactivation. It is also conceivable that insufficient spindle and/or joint mechanoreceptor feedback would have an effect on the function of the shoulder girdle muscles via direct or indirect reflex pathways (Jerosch *et al.*, 1993; Jerosch *et al.*, 1997; Voigt *et al.*, 1998). Such a process would affect the dynamic stability and neuromuscular control of the joint either through a failure to inhibit agonist muscles at certain points in the movement cycle (for e.g., when approaching the limits of the range of motion), or through a failure to stimulate the rotator cuff muscles to act at the appropriate time to maintain joint stability. The hypothesis of deficient proprioception, either through supraspinal motor control mechanisms or through spinal reflex pathways, is also consistent with the findings of the EMG study, which identified atypical patterns of shoulder muscle activation in subjects with MDI.

Although the preceding discussion presents a plausible explanation for the results found in this thesis, there are also several alternative explanations that can also account for the data. One is that the motor behaviour exhibited by subjects with MDI is directly related to the pathological mechanics of the joint, specifically abnormal glenohumeral and/or scapulothoracic motion caused by the lack of passive capsuloligamentous restraint. The presence of abnormal joint mechanics would potentially create a mismatch between the feedforward commands issued by the controller and the resulting output of the system, in that the subsequent consequences of the actions of the muscles on the mechanics of the joint would be inconsistent with the commands issued by the controller. As with the hypothesis of deficient proprioception, this problem would also create a situation in which the predicted sensory consequences would be inconsistent with the actual sensory feedback (i.e., the reafferent signal).

It is conceivable that over an extended period of time, the processes of deficient proprioception and/or pathological joint mechanics would lead to a re-training or adaptation of the internal forward model. That is to say, it is reasonable to expect that the forward model would attempt to make accurate motor predictions given the dysfunctional or suboptimal mechanical system of which it is a part, although this might be difficult given the inherent uncertainty of the system. It is also conceivable that such a process would have an effect on the inverse dynamics model, such that it would become trained to transform the error signal into motor commands that more appropriately match the pathological dynamics of the system. Such a process would ultimately lead to modifications in the feedforward motor commands sent to the muscles responsible for moving and stabilizing the shoulder, which is a potential explanation for the results

presented in this thesis, which may or may not involve deficient proprioception. However, it is also important to consider that, given the integrated nature of the neuromuscular control system and the somatosensory link between passive and dynamic stability, it is very unlikely that such a process involving pathological joint mechanics would have no affect on the quality or quantity of joint proprioceptive feedback.

Another possibility, with respect to the sensorimotor control mechanisms that could be involved in MDI, is that chronic changes occur in the internal forward model, such that it adapts to make predictions about the sensory consequences of actions that involve pain and/or joint instability. It is reasonable to suggest that over an extended period of time the system would adapt to the instability (and/or the frequent episodes of pain and discomfort that accompany the instability), such that it develops a protective mechanism whereby the feedforward motor commands and associated muscle activation patterns are altered in some respect. In relation to this idea, and in the context of optimal control theory, it is also possible that the system adopts a different set of criteria with which to optimize performance. It is conceivable that the maintenance of joint stability is an over-riding concern to the extent that the presence of altered joint mechanics and/or deficient proprioceptive feedback dictates the selection of a different control strategy, one that selects a cost function that attempts to minimize the pain and/or joint instability.

In the specific case of MDI, the fact that capsuloligamentous laxity is not always synonymous with glenohumeral instability, argues to some degree against the pathological mechanics hypothesis, at least in its most exclusive form, which takes no account of the potential effects on sensory feedback mechanisms. It could be argued that because any passive changes in joint mechanics caused by glenohumeral hyperlaxity

should be present in a shoulder without instability, something other than laxity is necessary to cause MDI. As suggested by Lephart and Henry (1996), a deficiency in proprioceptive feedback most likely exacerbates an existing lack of passive stability, such that this in combination with decreased neuromuscular control is sufficient to generate the processes that lead to joint instability. In the control systems model, both deficient proprioceptive feedback and pathological joint mechanics would compromise sensory reafference, leading to processes that would reduce the fidelity of the motor control of the limb. Deficient proprioception would create a situation in which the resulting motor output is not adequately "recorded" by the somatosensory system, thus diminishing the quality of information used by the other processes in the system.

8.3 Effects of elastic shoulder bracing

The study of the effects of elastic bracing on dynamic upper limb proprioception demonstrated that wearing an elastic shoulder brace had no immediate effect on the reaching kinematics or hand trajectory endpoint position errors of subjects with MDI or of subjects with healthy shoulders. The results suggested that wearing an elastic shoulder brace provides no immediate functional benefit to the control of natural multijoint reaching movements of the type regularly experienced in everyday life (reaching to open doors, retrieving items off a shelf, etc.), which is an important concern for subjects with MDI. If an elastic brace is to confer any functional benefit to joint stability through the alleviation of deficient proprioception in subjects with MDI, then the brace needs to be able to facilitate proprioceptive input during the performance of natural, multijoint arm

movements. Consequently, this study used repetitive reaching movements, performed under no-vision conditions, to investigate the effect of the brace on the ability to use dynamic proprioception to reposition the endpoint of the limb in three-dimensional space.

The original intent of this study had been to assess both the immediate and long-term effects of elastic bracing on patients with MDI by performing a series of follow-up tests at regular intervals over a one-year period (3, 6 and 12 months). Although several re-tests were conducted, subject attrition and a lack of compliance in wearing the brace precluded any assessment of its long-term effects. Consequently, this study was limited to investigating the immediate effects of bracing on proprioceptive reaching task performance and did not address whether an elastic shoulder brace might provide functional motor performance benefits to individuals with MDI over an extended period of time. This question still needs to be addressed.

Preliminary clinical trial data collected by our research group has indicated that patients who regularly wear the brace experience improved recovery rates and reduction of symptoms compared to patients who do not. The data was collected as part of a clinical study on MDI, which was overseen and administered by Dr. R. Balyk and Ms. Charlene Luciak through the Northern Alberta Shoulder Program and included some of the same patients who were subjects in the experiments conducted for this thesis. The patients in the clinical study were assessed at regular intervals using standard physical examination procedures, including manual muscle strength and range of motion tests and the Western Ontario Shoulder Index (WOSI) clinical assessment tool. The discrepancy between the clinical results and the results of the reaching movement study most likely occurred because the clinical study assessed patients after they had worn the brace for an

extended period of time whereas this study did not. Although a number of studies in the literature have reported immediate improvements in joint proprioceptive acuity as a result of wearing an elastic brace, it is likely that the motor systems of healthy participants and the subjects with MDI would have required an extended period of time to adapt to any somatosensory changes induced by the brace. It is also possible that not detecting a difference between the braced and non-braced conditions was a consequence of using a protocol that measured the ability to use proprioception to guide and control the movement of the entire upper limb. A shoulder specific, passive joint-repositioning or threshold to detection of passive motion test might have been more applicable in detecting an immediate effect of the brace on proprioceptive acuity.

Similarly, it is also conceivable that elastic shoulder bracing might not have any effect on accuracy-constrained, low-speed, voluntary arm movements, but that for high-velocity ballistic actions or involuntary reflex responses the effect might be different. The sensorimotor control of limb movement and joint stability is intricately complex, and it may be that voluntary reaching movements do not place sufficient demands on the neuromuscular control system for an elastic brace to have any effect on upper limb motor performance.

8.4 Shoulder electromyography

The functional stability paradigm of Lephart & Henry (1996) proposed that decreased neuromuscular control occurs as a consequence of deficient joint proprioceptive feedback. In applying the precepts of this model to the shoulder, it is reasonable to infer

Appendix

Appendix I: Three-dimensional motion analysis system (Vicon) calibration

The procedure for the 3D Vicon motion analysis system calibration was completed as follows. Two retroreflective markers were attached to a Vernier caliper (the criterion), such that the distance between the markers could be precisely measured and varied. The measurement points of the caliper were aligned as closely as possible with the centroid of each marker. The caliper was oriented so that its length corresponded to one of the axes of the three-dimensional coordinate system (the x-plane). This was accomplished by using the Vicon system's calibration object to designate a reference point. The 4-camera system was instructed to collect data for a period of 5 seconds, which at a sampling rate of 60 Hz yielded 300 samples per trial. For the static trials, the calibration data were reconstructed using the same three-dimensional parameters (identical spatial volume) that were used in the target positioning error experiment.

For the dynamic trials, the reconstruction volume was increased by 500 mm in the plane of motion and the caliper was placed on a wheeled-cart that was pulled through a measurement zone of 3 metres. The average speed for the 7 trials (10 trials were conducted, however, the first 3 were discarded because of problems with securing the caliper) was 0.4 metres/second. The range of speeds (both within and between trials) closely approximated the reaching movement test trials. Figure 8.1 shows the results of one of the dynamic calibration trials, including the inter-marker distance (IMD), the mean value for the 300 samples and the ± 1 standard deviation for the trial. Overall, the calibration results exceeded expectations and the manufacturers reported resolution of 0.5 mm. This was likely the result of the relatively small calibration volume needed to

investigate reaching movements, compared with the larger volumes needed to investigate human locomotion.

Table 1: Static and Dynamic Vicon Calibration Trials

	Trial	Vernier Caliper Criterion (mm)	Vicon Mean (N = 300)	Standard Deviation	Absolute Difference (mm)
Dynamic Trials	4	110.0	110.597	0.065	0.60
	5	111.0	111.139	0.048	0.14
	6	120.0	119.765	0.041	0.23
	7	100.0	100.224	0.067	0.22
	8	101.0	101.276	0.058	0.28
	9	110.0	109.716	0.089	0.28
	10	105.0	105.080	0.047	0.08
Static Trials	1	104.0	103.936	0.056	0.06
	2	106.0	105.992	0.030	0.01
	3	130.0	130.000	0.049	0.00
	4	129.0	128.975	0.182	0.02
	5	131.0	130.906	0.082	0.09
	6	129.5	129.464	0.053	0.04
	7	129.6	129.574	0.043	0.03
	8	80.0	80.0885	0.046	0.09
	9	79.0	79.0127	0.045	0.01
	10	78.0	77.8703	0.075	0.13

Mean Dynamic Trial Absolute Difference = 0.261 mm

Mean Static Trial Absolute Difference = 0.048 mm

Combined Mean Absolute Difference = 0.136 mm

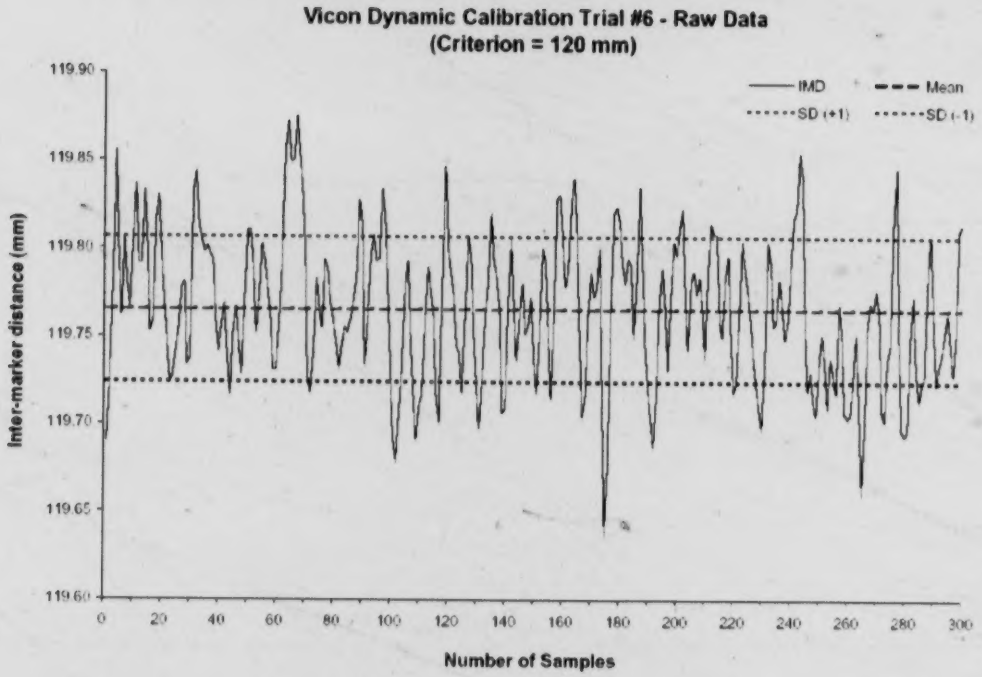

Figure 1: *Representative dynamic calibration trial for the Vicon 140 three-dimensional motion analysis system. The criterion (Vernier caliper) for this particular trial was 120.0 mm. The Vicon 140 mean reconstructed inter-marker distance was 119.765 mm.*

Appendix II: Statistical analysis

The primary factor for comparison in these experiments was the subject group; i.e., the comparison between subjects with MDI and control subjects with normal, healthy shoulders. This was done in all cases except for the first paper on cyclic pointing movements and the fourth paper on bracing. In the experiments conducted for this thesis, several other within-group factors were assessed, among them limb type and movement

type. Because repetitive test procedures were used to acquire the data, repeated measurements of the dependent variables were also obtained. Consequently, these experiments possessed a between-within design. That is to say, the analysis of data in all cases, except for the first paper (Ch. 3) and the bracing study, required the inclusion of both between-groups and within-groups (repeated measures) factors. As a result, two-way and three-way repeated measures analyses of variance were used to statistically analyze the data.

Appendix III: Subject information sheets and consent forms

See following page.

Proprioceptive Deficiencies in Individuals with Multidirectional Shoulder Instability (MDSI)

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PURPOSE OF THE STUDY: We want to know if people with unstable shoulders have problems sensing arm position. We also want to find out if this ability can be improved using a custom shoulder brace.

BACKGROUND: Specific muscles contribute to shoulder stability. Sensation of joint position is important for controlling these muscles. Injury to the structures providing this information may result in an unstable shoulder. It is thought that joint position sense deficits exist in patients with unstable shoulders. We will examine whether this sense can be improved with a shoulder brace.

PROCEDURE: Small round balls ("joint markers") will be attached to your arms. Eight joint markers will be used. Video cameras will be used to track the marker positions. You will be required to reach forwards and upwards. These movements will be performed from a seated position. You will reach towards a target. The target will be placed in front of you and above you. You will be shown how to perform these movements (reaching forward and reaching overhead) correctly. You will also be given verbal instructions on how to do this. You will have two practice trials to become familiar with each test movement. The goal of both movements is to position your finger as closely as possible to the target. Then you will be blindfolded and allowed two more practice trials. Your goal is to mimic the visual trials as closely as possible. Following the four practice trials, you will perform the movement ten more times. Ten trials of both movements, for both arms, will be performed. You will also be required to repeat the above process while wearing an upper body brace (orthosis). This process will take approximately one hour to complete.

RISKS/BENEFITS: There are no known risks to participating in this study. Your involvement will help us to understand the problem of shoulder instability. A shoulder brace may provide patients with a new treatment for this type of shoulder problem.

CONFIDENTIALITY/FREEDOM TO WITHDRAW: Data will be identified by code. This code will include your first and last initials. Your name and other personal information will not be made public. Your decision to help in this study is voluntary. You are free to withdraw at any time. This will not affect the treatment provided to you by your physician, the hospital or its staff. It may be necessary for Dr. Balyk to refer to patient charts for information regarding age, diagnosis and/or treatment. Data from this experiment will not be used in future studies without further approval from an ethics committee.

If you have questions about the study please contact Mr. Jim Raso. He can be reached at 471-2262, ext. 2390. For comments about the research procedures contact Dr. K. Pain. Dr. Pain is the Director of Research Services and can be reached at 471-8212.

Please initial below to indicate that you have read and understand the information described on this page.

Initial of Research Participant _____

Initial of Investigator _____

CONSENT FORM

Proprioceptive Deficiencies in Individuals with Multidirectional Shoulder Instability (MDSI)

Principal Investigator(s): **Dr. R. Balyk , Department of Surgery, 439-5858, Fax: 433-1344**

Co-Investigator(s): **John M. Barden**

Part 2 (to be completed by the research subject):

Do you understand that you have been asked to be in a research study?	Yes	No
Have you read and received a copy of the attached Information Sheet?	Yes	No
Do you understand the benefits and risks involved in taking part in this research study?	Yes	No
Have you had an opportunity to ask questions and discuss this study?	Yes	No
Do you understand that you are free to refuse to participate or withdraw from the study at any time? You do not have to give a reason and it will not affect your care.	Yes	No
Has the issue of confidentiality been explained to you? Do you understand who will have access to your records?	Yes	No
Do you want the investigator(s) to inform your family doctor that you are participating in this research study? If so, please provide your doctor's name: _____	Yes	No

This study was explained to me by: _____

I agree to take part in this study.

_____ _____ _____

Signature of Research Participant Date Witness

_____ _____

Printed Name Printed Name

I believe that the person signing this form understands what is involved in the study and voluntarily agrees to participate.

_____ _____

Signature of Investigator or Designee Date

THE INFORMATION SHEET MUST BE ATTACHED TO THIS CONSENT FORM AND A COPY GIVEN TO THE RESEARCH SUBJECT

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Clinical Neurophysiology 116 (2005) 1846–1857

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Atypical shoulder muscle activation in multidirectional instability

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Accepted 28 April 2005
Available online 27 June 2005

Abstract

Objective: Surface and intramuscular electromyography was used to investigate shoulder muscle activity in subjects with multidirectional instability (MDI).

Methods: Subjects (seven MDI, 11 control) performed repetitive shoulder abduction/adduction, flexion/extension and internal/external rotation movements on an isokinetic dynamometer. The activity of the deltoid, infraspinatus, supraspinatus, latissimus dorsi, and pectoralis major muscles were recorded using double-differential surface and intramuscular fine-wire electrodes. A repeated measures analysis of variance evaluated group differences in the amplitude, onset, termination and duration of the muscle activity.

Results: Significant activation parameter differences for the supraspinatus, infraspinatus, posterior deltoid and pectoralis major muscles were found in the subjects with MDI. The rotator cuff and posterior deltoid muscles demonstrated abbreviated periods of activity when performing internal/external rotation, despite activation amplitudes that were similar to the controls. In contrast, the activation of the pectoralis major differed from the control group in both the amplitude and time domains when performing shoulder extension.

Conclusions: MDI is associated with atypical patterns of muscle activity that occur even when highly constrained movements are used to elicit the activity.

Significance: In addition to glenohumeral hyperlaxity, the results suggest that dysfunctional neuromuscular control of the rotator cuff is also a contributing factor to the pathoetiology of MDI.

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Keywords: Multidirectional shoulder instability; Electromyography; Temporal activation characteristics; Rotator cuff

1. Introduction

The shoulder girdle is a complex, multi-articulated functional unit capable of achieving great mobility. This mobility arises through the coordinated activation of numerous muscles, which act to simultaneously produce limb segment motion and maintain joint stability. Along with active neuromuscular control, the functional integrity of the glenohumeral joint is maintained through a variety of passive mechanisms, including negative intra-articular pressure, bony restraint and capsuloligamentous tension (Bigliani et al., 1996; Hurschler et al., 2000; Jerosch et al., 1996; Matsen III, 1994). Of these mechanisms, only the neurally mediated control of muscle activation has the capacity to adapt joint mechanics in response to different limb configurations and loading conditions. Studies indicate that a failure of this adaptive mechanism is associated with the incidence of glenohumeral instability (Glousman et al., 1988; Kronberg et al., 1991; McMahon et al., 1996). Consequently, the investigation of atypical shoulder muscle function is essential to provide a complete understanding of the pathomechanics of glenohumeral instability and to develop more effective treatment strategies for shoulder instability related conditions.

One of these conditions is multidirectional instability (MDI), which is characterized by symptomatic subluxation and/or dislocation of the glenohumeral joint in more than

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doi:10.1016/j.clinph.2005.04.019

Abstract

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Conclusions: MDI is associated with atypical patterns of muscle activity that occur even when highly constrained movements are used to elicit the activity.

Significance: In addition to glenohumeral hyperlaxity, the results suggest that dysfunctional neuromuscular control of the rotator cuff is also a contributing factor to the pathoetiology of MDI.

7.1 Introduction

The shoulder girdle is a complex, multi-articulated functional unit capable of achieving great mobility. This mobility arises through the coordinated activation of numerous muscles, which act to simultaneously produce limb segment motion and maintain joint stability. Along with active neuromuscular control, the functional integrity of the glenohumeral joint is maintained through a variety of passive mechanisms, including negative intra-articular pressure, bony restraint and capsuloligamentous tension (Bigliani *et al.*, 1996; Hurschler *et al.*, 2000; Jerosch *et al.*, 1996; Matsen III, 1994). Of these mechanisms, only the neurally mediated control of muscle activation has the capacity to adapt joint mechanics in response to different limb configurations and loading conditions. Studies indicate that a failure of this adaptive mechanism is associated with the incidence of glenohumeral instability (Glousman *et al.*, 1988; Kronberg *et al.*, 1991; McMahon *et al.*, 1996). Consequently, the investigation of atypical shoulder muscle function is essential to provide a complete understanding of the pathomechanics of glenohumeral instability and to develop more effective treatment strategies for shoulder instability related conditions.

One of these conditions is multidirectional instability (MDI), which is characterized by symptomatic subluxation and/or dislocation of the glenohumeral joint in more than one direction. The pathology of MDI typically involves a loose and redundant joint capsule with episodes of recurrent pain and instability, particularly in the midrange positions of glenohumeral motion (An & Friedman, 2000; Arendt, 1988; Beasley *et al.*, 2000; Schenk & Brems, 1998). Midrange stability requires the precise activation and

control of the rotator cuff muscles, more so than end range stability, which depends to a greater extent on passive ligament tension (Bigliani *et al.*, 1996; Janwantanakul *et al.*, 2001; Mallon & Speer, 1995; Nyland *et al.*, 1998). Not surprisingly, rotator cuff dysfunction has been implicated as a contributing factor to the development of MDI (Mallon & Speer, 1995; Schenk & Brems, 1998). While it is reasonable to suggest that deficient rotator cuff activation compounds the lack of stability created by excessive capsuloligamentous laxity, no experimental evidence currently exists to support this hypothesis.

The relationship between glenohumeral instability and shoulder muscle function has been investigated in voluntary posterior instability (Beall *et al.*, 1987; Pande *et al.*, 1989), recurrent anterior instability (Glousman *et al.*, 1988; Kronberg *et al.*, 1991; McMahon *et al.*, 1996; Wallace *et al.*, 1997) and most recently, MDI (Morris *et al.*, 2004). In each of these studies, subjects with instability were found to have movement specific, abnormal levels of activation in several different shoulder muscles. For subjects with MDI, Morris *et al.* (2004) investigated six different shoulder muscles and found atypical levels of activation for all three sections of the deltoid, but found no differences in the activation of the rotator cuff. Electromyographic investigations of glenohumeral instability have quantified the level of muscle recruitment by comparing the amplitude of activation at different stages of the movement cycle. While recruitment level is an important characteristic of muscle function, an equally important factor is the timing of the activation. To our knowledge, a detailed analysis of the temporal characteristics of shoulder muscle activation in MDI or any other type of glenohumeral instability has not been conducted. The purpose of this study, therefore, was to use electromyography

(EMG) to investigate the timing and amplitude of shoulder muscle activation in subjects with MDI. We hypothesized that the shoulder muscles of subjects with MDI, and in particular the rotator cuff, would demonstrate temporal activation characteristics that were different from control subjects with normal, healthy shoulders.

7.2 Materials and Methods

Subjects

Seven male subjects with MDI (age = 25.0 ± 9.5 years) and 11 male subjects with normal, healthy shoulders (age = 29.6 ± 9.7 years) participated in the study. The diagnosis of MDI was made on the basis of a comprehensive physical examination conducted by an experienced orthopaedic shoulder specialist (R.B.). The examination consisted of inspection, palpation, range of motion, and instability testing performed on both the symptomatic and asymptomatic shoulders (Arendt, 1988; Schenk & Brems, 1998). The tests for instability included the load and shift test, the sulcus test and the apprehension test (Arendt, 1988; Mallon & Speer, 1995; Schenk & Brems, 1998). Subjects in the MDI group possessed the following characteristic features of MDI: (1) a functionally significant inability to keep the humeral head centered in the glenoid fossa, especially in positions not at the extremes of motion; (2) the absence of an injury mechanism likely to tear the glenohumeral ligaments; (3) spontaneous reductions of translations; (4) glenohumeral translations that duplicated the symptoms of concern to the patient; (5) a diminished resistance to translation in multiple directions as compared with a normal glenohumeral joint; and (6) an absence of traumatic lesions (Matsen III, 1994).

Subjects in the MDI group also possessed no confounding genetic or neurological health conditions (for example, Ehlers-Danlos syndrome or syringomyelia) and received no treatment or intervention prior to testing. Control subjects had no history of shoulder injuries, complaints or surgery. Subjects in both groups provided informed consent prior to participating in the study, which was approved by the author's local institutional Ethics Review Board.

Electromyography

Double-differential surface electrodes were used to record the electrical activity of the anterior, middle and posterior fibres of the deltoid, the latissimus dorsi and the pectoralis major (sternal head). Bipolar intramuscular fine-wire electrodes were used to record the activity of the infraspinatus and supraspinatus. The double-differential surface electrode configuration (3 electrodes instead of 2) was used as a precautionary measure to increase the selectivity of the electrode detection area, thus reducing the potential for crosstalk contamination from neighbouring muscles (DeLuca & Merletti, 1988; Koh & Grabiner, 1993; Solomonow *et al.*, 1994; Winter *et al.*, 1994). Two types of disposable, pre-gelled ECG surface electrodes were used: Red Dot neonatal 2258 electrodes (3M, St. Paul, MN) for the anterior, middle and posterior deltoid and MediTrace 310781 electrodes (Kendall-LTP, Chicopee, MA) for the latissimus dorsi and pectoralis major. The slightly smaller neonatal electrodes (3.2 cm in diameter vs. 3.65 cm) were specifically selected for the deltoid because their reduced size permitted greater separation between the electrode groupings for the different sections of the muscle. Following skin preparation, the electrodes were positioned parallel to the direction of the muscle fibres according to

Table 7.2: Control and MDI Group Mean (SD) EMG Parameters for the Latissimus Dorsi and Pectoralis Major

		Latissimus Dorsi				Pectoralis Major			
		AMP (%)	ONST (°)	TERM (°)	ROM (°)	AMP (%)	ONST (°)	TERM (°)	ROM (°)
Abduction	Control	49.7 (15.7)	2.7 (8.5)	-40.1 (49.8)	127.4 (50.7)	–	–	–	–
	MDI	59.2 (22.9)	7.2 (18.8)	-9.7 (59.6)	92.5 (57.5)	–	–	–	–
Adduction	Control	51.8 (13.0)	-8.5 (9.1)	21.7 (28.7)	76.9 (28.3)	51.3 (11.7)	-5.7 (4.2)	21.1 (18.9)	74.6 (20.7)
	MDI	52.6 (10.8)	-7.6 (8.3)	26.0 (24.5)	72.0 (26.4)	46.6 (11.4)	-5.3 (8.5)	18.3 (45.8)	76.9 (50.1)
Flexion	Control	53.1 (11.5)	-5.0 (11.2)	-18.5 (28.2)	113.4 (34.4)	53.7 (16.7)	-7.4 (10.5)	13.0 (58.3)	84.4 (62.3)
	MDI	52.9 (11.0)	-6.9 (12.1)	-20.2 (27.3)	117.1 (34.5)	50.5 (20.8)	-10.6 (11.2)	-2.0 (53.6)	102.6 (59.8)
Extension	Control	–	–	–	–	62.6 (16.2)*	-2.0 (1.8)	32.9 (41.7)†	59.6 (41.6)†
	MDI	–	–	–	–	37.3 (8.0)	-5.2 (3.8)	-29.2 (38.8)	124.4 (40.3)
Internal Rotation	Control	59.6 (8.0)	0.0 (1.9)	-0.1 (38.1)	90.1 (37.8)	46.6 (12.6)	-3.7 (2.9)	23.1 (36.5)	70.6 (37.7)
	MDI	55.8 (22.2)	-0.2 (5.1)	-18.3 (65.0)	107.7 (62.9)	45.3 (7.6)	-4.7 (5.2)	17.8 (18.4)	76.9 (20.5)
External Rotation	Control	60.0 (12.4)	-4.8 (21.0)	7.5 (20.9)	87.3 (29.6)	–	–	–	–
	MDI	57.0 (9.9)	-16.9 (25.5)	-4.9 (34.2)	111.8 (45.4)	–	–	–	–

*The * indicates a significant difference of $p < 0.005$ between the control and MDI groups, while the † indicates a difference of $p < 0.01$.*

AMP or ONST parameters. However, significant group main effects were found for the TERM [F (1, 48) = 4.55, $p < 0.05$] and ROM [F (1, 48) = 3.99, $p < 0.05$] parameters. The significant main effect for TERM occurred because the deactivation of the supraspinatus muscle in the MDI group preceded that of the controls for all three movement conditions. Post-hoc testing for TERM demonstrated that none of the MDI vs. control comparisons were significant, however, the difference for external rotation approached significance at $p < 0.09$. With respect to the significant group main effect for the ROM parameter, the data in Figures 7.2 to 7.4 indicate that the contraction duration of the supraspinatus was less for the MDI group than it was for the control group. Post-hoc tests revealed that the abbreviated supraspinatus activation was significant ($p < 0.01$) during external rotation (Figure 7.4).

Infraspinatus

As with the supraspinatus, the group results for the infraspinatus during shoulder abduction, flexion and external rotation are presented in Figures 7.2, 7.3 and 7.4, respectively. The statistical analyses for the four EMG parameters demonstrated that there were no significant between-group effects for the AMP parameter. For the ONST parameter, a significant main effect for task was identified [F (2, 47) = 3.47, $p < 0.05$], such that the ONST for shoulder flexion preceded that for external rotation in both groups ($p < 0.05$). Similarly, a significant main effect for task was identified for the TERM parameter [F (2, 47) = 5.37, $p < 0.01$]. The data show that for shoulder abduction (Figure 7.2), the termination of the infraspinatus occurred before the end of the abduction phase (+ mean values for TERM), whereas for shoulder flexion (Figure 7.3) the infraspinatus

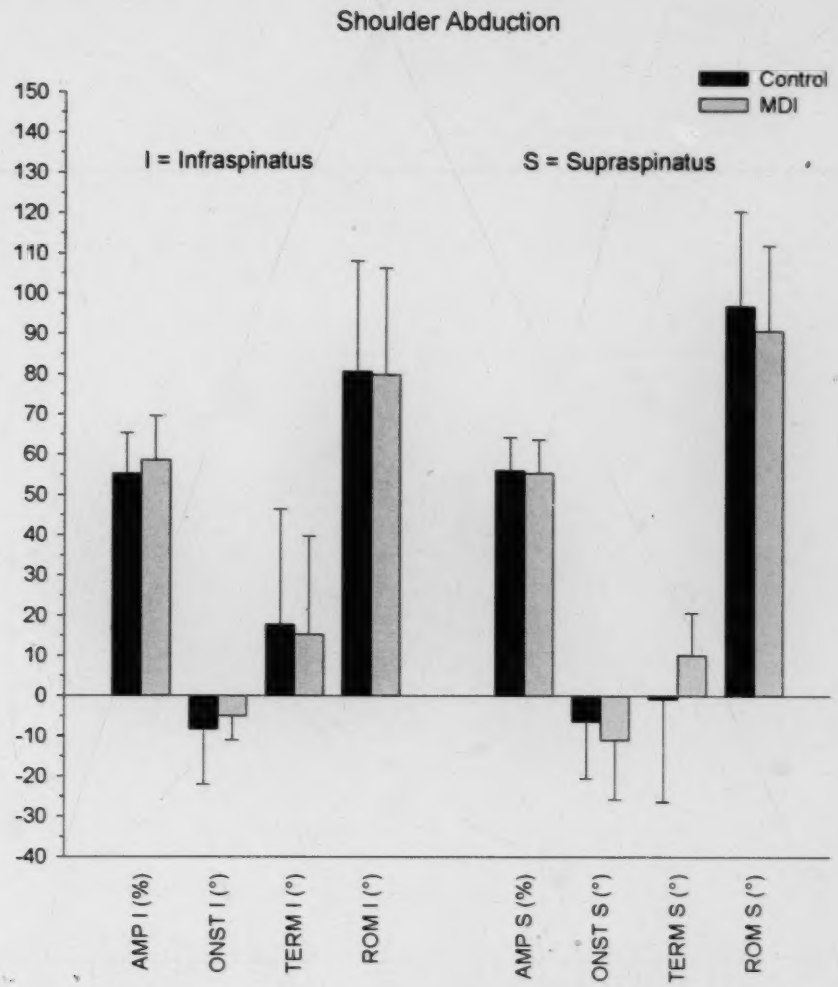

Figure 7.2: *Group mean activation parameter values for the infraspinatus (I) and supraspinatus (S) muscles during shoulder abduction. AMP (%) = normalized (% peak EMG) average amplitude; ONST (°) = onset of activation; TERM (°) = termination of activation; ROM (°) = activation range of motion. Error bars = standard deviation of the mean.*

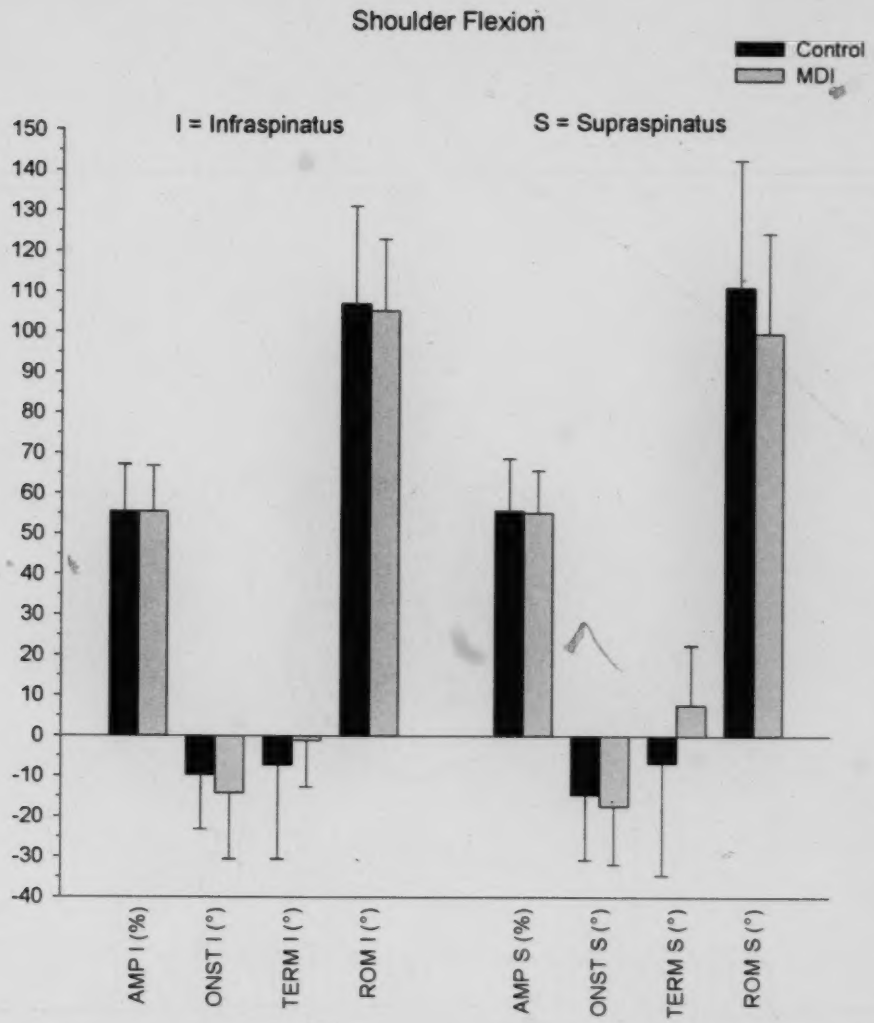

Figure 7.3: *Group mean activation parameter values for the infraspinatus (I) and supraspinatus (S) muscles during shoulder flexion. Error bars = standard deviation of the mean.*

was deactivated shortly after the reversal from flexion to extension (the negative values demonstrate that although the muscle was active during flexion, the activation terminated

at the beginning of the extension phase). Post-hoc testing confirmed that the difference in the termination angle between shoulder abduction and flexion was significant ($p < 0.01$).

As with the ONST and TERM parameters, a significant main effect for task was also found for the ROM parameter [F (2, 47) = 8.62, $p < 0.001$]. Post-hoc testing indicated that the ROM for both groups was significantly greater for shoulder flexion than for shoulder abduction ($p < 0.001$) and external rotation ($p < 0.05$). In the course of conducting the pair-wise comparisons to identify the inter-task differences for the ROM parameter, it was found that the difference between the MDI group and the control group for external rotation (Figure 7.4) was significant ($p < 0.01$), such that subjects with MDI activated the infraspinatus throughout a decreased range of motion relative to the controls. This reduced contraction duration occurred because subjects in the MDI group activated and deactivated the infraspinatus later and earlier, respectively, relative to subjects in the control group (Figure 7.4).

Posterior Deltoid

For the posterior deltoid, the statistical analysis indicated that there were no significant differences for the normalized average amplitude (AMP) parameter, while for the onset of activation (ONST), significant group [F (1, 48) = 3.99, $p < 0.05$] and group x task interaction effects [F (2, 48) = 8.50, $p < 0.001$] were observed. Post-hoc analyses revealed that the onset of activation for subjects with MDI was significantly ($p < 0.05$) delayed relative to the control group for external rotation (see Table 7.1). The data in Table 7.1 demonstrate that for all movement conditions, the ONST of activity for both groups occurred prior to the movement phase in which the muscle was most active (as

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8

General discussion, conclusions and recommendations

8.1 Introduction

The purpose of this thesis was to investigate two hypotheses that have been proposed in the literature to account for the development of idiopathic MDI. The first hypothesis investigated was that MDI is associated with a proprioceptive deficiency of the symptomatic shoulder and the second was that the presence of atypical shoulder muscle activation, particularly in the rotator cuff, is responsible for the lack of dynamic stability that occurs in this condition. To investigate the first hypothesis, the fidelity of goal-directed reaching movements to remembered proprioceptive targets was investigated in subjects with healthy shoulders and in subjects with MDI. In relation to the hypothesis of deficient proprioception, subjects with MDI were also tested with and without an elastic shoulder brace to determine whether the brace was able to provide any immediate benefit to using proprioception to control the positioning of the arm when performing repetitive goal-directed tasks. To test the hypothesis related to dysfunctional shoulder muscle activity, surface and fine-wire electromyography was used to record the activity of the supraspinatus, infraspinatus, deltoid, latissimus dorsi and pectoralis major muscles in subjects with MDI and in control subjects with healthy shoulders, during the performance of controlled planar shoulder movements performed on an isokinetic dynamometer.

In this chapter, a brief discussion and review of the findings of each paper is presented. The first section discusses the general results of the repetitive proprioceptive reaching task experiments presented in Chapters 3, 4 and 5. The next section discusses the results of the elastic bracing study presented in Chapter 6. The final section provides a

general discussion of the shoulder EMG study presented in Chapter 7. Following these discussions, several recommendations for future research directions are presented.

8.2 Repetitive reaching to remembered proprioceptive targets

In subjects with normal healthy shoulders, several new and interesting findings were presented that extended previous work on non-repetitive (Adamovich *et al.*, 1998; Adamovich *et al.*, 1999) and two-dimensional planar reaching movements (Brown *et al.*, 2003a; Brown *et al.*, 2003b) to remembered proprioceptive targets. The research in this thesis contributed to the literature by investigating repetitive, multijoint proprioceptive reaching tasks that occurred in three-dimensional space. It was found that hand trajectory position errors relative to the target and first movement trajectory accumulated as additional repetitions of the task were completed, which supports the recent work of Brown *et al.* (2003a) who also demonstrated hand trajectory drift for reaching to remembered proprioceptive targets. It was also found that movement extent was consistently underestimated throughout the trial, such that by the 10^{th} repetition it had decreased significantly relative to the amplitude of the first trajectory. Although the absolute spatial position errors and underestimation of movement extent consistently accumulated with repeated task performance, the movement reproduction capability, as determined by the hand trajectory inter-repetition error, improved relative to the initial and second movement reproductions. This finding, which suggests an updating of the sensorimotor control system, was consistent with the hand velocity profile kinematic

use of intramuscular fine-wire electrodes (inserted with hypodermic needles) was a detriment to subject participation. This was more of a limiting factor for the MDI group than for the control group, given that the population of subjects with MDI is considerably smaller. Numerous technical obstacles and challenges were also encountered, the most significant of which concerned the amount of work required to design, manufacture and test the double-differential surface electrodes.

8.5 Summary and future recommendations

The research conducted for this thesis has made an important contribution to the development of knowledge in several fields of study. First, there are only a small number of studies that have investigated deficient proprioception in shoulder instability, and none have investigated this relationship as it applies to MDI. Numerous review papers describe the clinical evaluation, diagnosis and pathophysiology of MDI but little exists in the way of controlled experimental investigation. Consequently, it is likely that the research presented in this thesis represents the single largest contribution to the investigation of factors that contribute to the development of MDI. Second, this thesis has contributed new information to the field of study concerned with the neural control of reaching movements. In particular, the motor behaviour associated with using proprioception to perform repetitive, three-dimensional reaching movements had not been investigated. Similarly, the effect of elastic bracing on dynamic upper limb proprioception, and the temporal characteristics of shoulder girdle muscle activity, had not been investigated in any type of shoulder instability. As a consequence of the research conducted for this

thesis, several recommendations for investigating the relationship between deficient proprioception and shoulder instability can be made.

First, an increase in the number of studies is necessary and these studies should attempt to test larger samples of subjects. Although the numbers of subjects used in the current experiments were consistent with those reported in the literature, it is possible that the small sample sizes could have introduced Type II errors as a result of the low statistical power (e.g., the negative results of the bracing study). It is possible that the immediate effect of an elastic brace on the motor performance of the upper limb is too small to be detected but without a greater number of subjects this cannot be determined. Given that the recruitment of subjects is contingent on the researcher having access to patients, future studies should attempt to include several practicing physicians to maximize patient referrals. One way to increase the sample size in this way would be to conduct a multi-centre study, in which standardized test procedures are administered to different groups of subjects.

Second, a joint kinematic and kinetic analysis of repetitive reaching movements to remembered proprioceptive targets should be conducted for subjects with normal, healthy shoulders and for subjects with shoulder instability (and/or MDI). This would provide additional information about the relationship between dynamic proprioception and the joint mechanics of goal-directed reaching movements, as well as the mechanics of glenohumeral instability.

A third recommendation is to investigate the joint position sense of the glenohumeral joint in both the symptomatic and asymptomatic limbs of subjects with MDI using a passive joint repositioning test. Similarly, dynamic proprioception of the

asymptomatic limbs of subjects with unilateral MDI should also be investigated further. A passive joint repositioning study would provide additional information with respect to the hypothesis of deficient proprioception in MDI because the data would be specific to the glenohumeral joint, whereas further investigation of the contralateral uninvolved limbs with asymptomatic shoulders would help to determine the relationship between deficient proprioception and the etiology of MDI (i.e., whether deficient proprioception precedes or occurs after the development of MDI). Answers to these questions might lead to further development in understanding the mechanisms of joint instability, with potential implications for the treatment of other joint instability related conditions.

The effects of elastic bracing on passive shoulder and dynamic upper limb proprioception should also be investigated in subjects with MDI and other types of glenohumeral instability. Although the present study found that no immediate functional motor performance benefits were associated with wearing an elastic shoulder brace when reaching to remembered proprioceptive targets, it is important to investigate the long-term effects of wearing an elastic brace. In addition, the relationship between elastic bracing and conservative treatment should also be studied, in that bracing might serve to facilitate the benefits provided by neuromuscular training and/or shoulder muscle strengthening exercises. Anecdotal evidence suggests that this is the case, and that bracing leads to improvements in manual muscle strength tests for the affected limb in patients with MDI. Consequently, clinical trials should be conducted simultaneously with controlled scientific studies to assess the effects of elastic bracing on shoulder proprioception and glenohumeral instability.

Finally, further study is needed to determine the activation patterns of other shoulder girdle muscles in subjects with MDI, particularly the subscapularis and scapulothoracic muscles. The activation characteristics of these and other muscles should also be examined for different types of movements, particularly for unconstrained, three-dimensional, multijoint movements of the type investigated in the first part of the thesis. Similarly, quantitative EMG analysis should be used to provide a better understanding of the short and long-term effects of proprioceptive training and shoulder muscle strengthening exercises on the neuromuscular control of the rotator cuff, in patients with shoulder instability and with MDI. These procedures could also be used to investigate the long-term effects of elastic bracing on the neuromuscular control of muscles that act to control and stabilize the glenohumeral joint. This research would ultimately provide important new information about the effects of conservative therapy on the function of specific shoulder girdle muscles, which should lead to the development of more effective rehabilitation programs for the treatment of MDI and other forms of glenohumeral instability.

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Appendix IV: Human Movement Science paper

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Repetitive pointing to remembered proprioceptive targets improves 3D hand positioning accuracy

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Available online 3 June 2005

Abstract

Repetitive pointing movements to remembered proprioceptive targets were investigated to determine whether dynamic proprioception could be used to modify the initial sensorimotor conditions associated with an active definition of the target position. Twelve blindfolded subjects used proprioception to reproduce a self-selected target position as accurately as possible. Ten repetitions for each limb were completed using overhead and scapular plane pointing tasks. A 3D optical tracking system determined hand trajectory start and endpoint positions for each repetition. These positions quantified three-dimensional pointing errors relative to the target position and the initial and preceding movement repetitions, as well as changes in movement direction and extent. Target position and cumulative start position errors were significantly greater than the corresponding preceding movement (inter-repetition) errors, and increased as the trial progressed. In contrast, hand trajectory start and endpoint inter-repetition errors decreased significantly with repeated task performance, as did movement extent, although it was consistently underestimated for each repetition. Pointing direction remained constant, except for the angle of elevation for scapular plane pointing.

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doi:10.1016/j.humov.2005.03.004

Appendix V: Clinical Orthopaedics and Related Research paper

Dynamic Upper Limb Proprioception in Multidirectional Shoulder Instability

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Abstract: The ability of subjects with multidirectional shoulder instability to use proprioception to complete a series of upper limb repositioning tasks was investigated. Twelve subjects with multidirectional instability and 12 control subjects were blindfolded and instructed to use proprioception to reproduce a self-selected target position as accurately as possible. Subjects completed 10 repetitions for each limb using three distinct upper limb movements: overhead reaching, scapular plane pointing, and humeral external rotation with abduction. A three-dimensional video motion analysis system tracked limb position and determined spatial hand position error. Subjects with multidirectional shoulder instability showed significantly greater hand position error than control subjects. No hand position error differences were found between the symptomatic and the asymptomatic limbs of subjects in the instability group. Inter-repetition error for subjects in both groups improved significantly during the first three movement cycles. These results suggest that after movement initiation, dynamic proprioception was a factor in improving hand position accuracy in both groups, but to a lesser degree in subjects with multidirectional instability. Consequently, subjects with multidirectional instability may have a reduced capacity to use proprioception to refine and control the motor output of the upper limb.

(*Clin Orthop* 2004;420:181–189)

Received for publication March 28, 2002; revised March 24, 2003; accepted May 6, 2003.

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Funding for this study was received from the Edmonton Orthopaedic Research Fund, and John M. Barden received additional funding from the Province of Alberta Graduate Fellowship program and a 75th Anniversary Scholarship Award from the Faculty of Medicine and Dentistry at the University of Alberta.

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DOI: 10.1097/01.blo.0000126220.04900.6e

Multidirectional instability of the shoulder is a complex condition that can be difficult to diagnose and to treat.[2,3,36,39] Neer and Foster first recognized multidirectional instability as a unique and separate condition from unidirectional instability, and developed the inferior capsular shift as a specific surgical procedure for its treatment.[44] Multidirectional instability can occur in males and females, in different age groups, and in most segments of the population, from sedentary individuals to elite athletes[3] and is considered to be a serious and more prevalent condition than previously realized.[2] It is characterized by symptomatic global laxity of the glenohumeral joint[6] and may present either traumatically, atraumatically, unilaterally, bilaterally, or with or without generalized joint laxity.[14,19,36,54] Individuals who possess multidirectional instability subluxate or dislocate anteriorly, posteriorly, or inferiorly with concurrent reproduction of symptoms in at least two directions.[6,54] Symptoms typically are associated with the midrange positions of glenohumeral motion and often occur during activities of daily living.[6,54]

The glenohumeral joint's relatively poor osseous and capsuloligamentous stability necessitates a reliance on neuromuscular proprioception and stabilization more than any other joint in the human body.[45] Proprioception is essential for coordination of multijoint movement sequences,[7,17,25,27] for regulating the amplitude and timing of antagonistic muscle activation for the purpose of limb segment deceleration,[22] and for regulating the contraction of distal segment muscles to control proximally produced joint interaction torques.[52,55] Proprioceptive deficiencies, which exist in patients with functional deafferentation, create major deficits in movement control[26,29,40,51–53] which result in increased movement variability,[40,53] the inability to maintain stable hand postures without visual guidance,[53] and a reduced capacity to detect and correct motion errors based on limb movement information after executing steady-state motor commands.[25,29,40,52,53] Studies of the relationship between proprioception and joint stability acknowledge the complementary role of sensorimotor systems and passive joint structures to the maintenance of functional joint stability.[5,31,32] Proprioception provides a mechanism by which the neuromuscular control system operates as a kind of

Appendix VI: Clinical Neurophysiology paper

Schenk, T. J., & Brems, J. J. (1998). Multidirectional instability of the shoulder: Pathophysiology, diagnosis and management. *J. Am. Acad. Orthop. Surg.*, **6**, 65-72.

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7

Atypical shoulder muscle activation in multidirectional instability

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Published in Clinical Neurophysiology 2005; 116: 1846-1857

termination (TERM). The raw electromyogram is depicted in (A) while (B) depicts the full-wave rectified signal after processing with the 50 millisecond moving average window. The time-based ONST and TERM values were reported as movement-based relative angular position values. The range of motion (ROM) parameter (Panel B) was determined by the ONST and TERM parameters. The lower dashed line in B indicates the 3 standard deviation (SD) criterion used to determine ONST and TERM. Note that the posterior deltoid was active primarily during abduction (Panel A), but that the onset of activation preceded the initiation of the abduction phase (i.e., ONST occurred approximately 7 degrees before the end of adduction). This demonstrates that prior to acting as an agonist to raise the arm, the deltoid was activated eccentrically to decelerate the arm at the end of adduction.

7.3 Results

Tables 7.1 and 7.2 summarize the group mean EMG parameter data for the anterior, middle and posterior deltoid (Table 7.1) and the latissimus dorsi and pectoralis major muscles (Table 7.2), respectively. The data for the infraspinatus and supraspinatus muscles are presented in Figures 7.2, 7.3 and 7.4. Several significant EMG parameter differences were found between the two groups, however, no significant within-group differences were found for any of the muscles tested. The absence of significant within-group effects demonstrates the inter-repetition consistency of the electromyograms for all of the movement conditions. Significant differences between the MDI group and the control group were found for the supraspinatus, infraspinatus, posterior deltoid and

Table 7.1: Control and MDI Group Mean (SD) EMG Parameters for the Anterior, Middle and Posterior Deltoid

		Anterior Deltoid		Middle Deltoid		Posterior Deltoid	
		Control	MDI	Control	MDI	Control	MDI
Abduction	AMP (%)	52.1 (7.4)	51.9 (11.6)	51.3 (8.6)	52.6 (8.0)	50.8 (7.3)	50.1 (10.2)
	ONST (°)	-6.1 (7.8)	-4.9 (7.0)	-10.1 (12.1)	-6.2 (3.5)	-5.5 (7.3)	-10.3 (17.0)
	TERM (°)	8.6 (8.6)	12.5 (26.3)	5.4 (6.0)	7.8 (5.7)	2.3 (4.4)	7.6 (13.8)
	ROM (°)	87.5 (11.5)	82.4 (14.2)	94.6 (14.3)	88.2 (4.7)	93.2 (8.4)	92.7 (27.4)
Flexion	AMP (%)	54.4 (7.6)	50.3 (6.7)	57.6 (10.1)	56.4 (7.4)	47.5 (9.3)	53.3 (8.1)
	ONST (°)	-15.9 (6.3)	-14.2 (12.4)	-3.8 (4.5)	-11.2 (14.8)	-3.3 (4.4)	-7.5 (7.9)
	TERM (°)	2.9 (29.7)	2.6 (2.0)	4.2 (4.2)	3.9 (3.2)	1.8 (3.3)	3.9 (3.4)
	ROM (°)	103.0 (13.3)	101.7 (12.2)	89.6 (5.8)	97.2 (14.5)	91.5 (5.4)	93.6 (9.0)
External Rotation	AMP (%)	57.2 (10.4)	52.0 (7.7)	53.2 (10.4)	48.0 (12.3)	42.9 (10.2)	49.5 (10.4)
	ONST (°)	-8.4 (24.8)	-13.9 (3.7)	-25.9 (20.9)	-32.9 (27.8)	-46.6 (25.8)*	-12.2 (19.4)
	TERM (°)	25.5 (2.6)	27.4 (7.3)	30.2 (21.5)	28.0 (38.1)	16.9 (21.7)	20.1 (24.7)
	ROM (°)	73.0 (30.8)	76.6 (36.0)	85.6 (28.4)	94.6 (38.7)	119.7 (28.5)	82.1 (40.5)

*The * indicates a significant difference of $p < 0.05$ between the control and MDI groups.*

pectoralis major muscles. No significant differences were found for the anterior deltoid, middle deltoid or latissimus dorsi.

Supraspinatus

The group mean parameter values for the supraspinatus are presented in Figures 7.2, 7.3 and 7.4. The statistical analyses identified no significant between-group effects for the

major well into flexion, and demonstrated normalized levels of activation that were 68% less than the control values.

Conclusion

This study identified atypical patterns of activation for the supraspinatus, infraspinatus, posterior deltoid and pectoralis major muscles of subjects with MDI, particularly when the shoulder was externally rotated in the abducted position. Several studies have shown that anterior instability (Lephart *et al.*, 1994; Warner *et al.*, 1996; Zuckerman *et al.*, 2003) and clinically determined generalized joint laxity (Blasier *et al.*, 1994) are associated with proprioceptive deficiencies of the shoulder. Recently, Barden *et al.* (2004) demonstrated that subjects with MDI are unable to use dynamic proprioception to reposition the arm as accurately as control subjects with normal, healthy shoulders. Given that excessive capsular laxity is one of the characteristic features of MDI, it is conceivable that a lack of tension in the glenohumeral ligaments and surrounding joint capsule could precipitate a decreased stimulation of local proprioceptive afferents in the shoulder. A deficiency in the capacity to generate proprioceptive feedback would provide a potential mechanism for how atypical shoulder muscle activity might occur in MDI. The results of the current study also suggest that neuromuscular training exercises for the rotator cuff (Kiss *et al.*, 2001; Lephart & Henry, 1996; Swanik *et al.*, 2002) should be routinely incorporated into the physical therapy programs typically prescribed for patients with MDI. Future studies should use EMG to determine the functional role and/or dysfunction of individual shoulder muscles for an increased number of constrained and unconstrained movement

conditions. This should lead to an improved understanding of the complex pathoetiology of multidirectional shoulder instability.

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that if individuals with MDI have deficient proprioception, then the muscles responsible for stabilizing and controlling the motion of the glenohumeral joint (particularly the rotator cuff muscles) should demonstrate atypical patterns of activation compared to the muscles of normal, healthy shoulders. The shoulder EMG study investigated the activation of five different shoulder muscles for a number of constrained, planar shoulder movements performed on an isokinetic dynamometer, and found differences in the timing of the pectoralis major, posterior deltoid, infraspinatus and supraspinatus activation in subjects with MDI. As discussed in section 8.2, a proprioceptive deficiency is not the only mechanism that could cause dysfunctional shoulder muscle activity. Maladapted feedforward motor commands resulting from pathological glenohumeral joint mechanics could also alter the activation patterns of the rotator cuff and other shoulder girdle muscles. Nevertheless, although the findings do not provide direct evidence of a proprioceptive deficiency in the symptomatic shoulders of subjects with MDI, they are consistent with one of the most important consequences predicted by this hypothesis, namely dysfunctional neuromuscular control of the rotator cuff.

Previous studies of muscle function in shoulder instability have used identical methods of analysis, in which the electromyographic signals were processed using a linear envelope and normalized over the entire range of motion to quantify the average levels of activity at different points in the movement cycle (Glousman *et al.*, 1988; Kronberg *et al.*, 1991; McMahon *et al.*, 1996; Morris *et al.*, 2004; Wallace *et al.*, 1997). While the levels of muscle activation in subjects with shoulder instability have been investigated, the temporal characteristics of the activation have not. Consequently, one of the objectives of this study was to quantify the temporal characteristics of shoulder

muscle activation in subjects with MDI. A second objective was to investigate the activity of the rotator cuff muscles in subjects with MDI, given that these muscles are the primary stabilizers of the glenohumeral joint. Unfortunately, the subscapularis was not included in the current analysis because the complexities and significant risks (lung puncture leading to pneumothorax) associated with inserting fine-wire electrodes into this muscle precluded its investigation. While this study attempted to analyze the principal muscles involved in moving and stabilizing the shoulder for each test movement, several factors limited the number of muscles that could be investigated. First, the subject preparation and test time was considerable (approximately two hours), particularly because of the extensive procedure required to insert the fine-wire electrodes. Although not extreme, this time commitment negatively affected subject participation, particularly as it pertained to taking time off work. Second, certain muscles were omitted because of methodological concerns or because they made a minor contribution to the movements in question (for example, the biceps brachii). The trapezius, for example, was not included in this study because electromyographic recordings of this muscle have been shown to be associated with low reproducibility and high inter-subject variability (Jensen *et al.*, 1993; Veiersted, 1991; Westgaard, 1988), in part because its complexity (multiple functional compartments) results in significant EMG variability in response to minor changes in electrode position.

While it is important to continue to investigate shoulder girdle muscle activity in subjects with MDI (particularly the activity of the subscapularis and scapulothoracic muscles), there are considerable limitations and difficulties associated with conducting this type of research, one of which involves subject recruitment. In the present case, the

INFORMATION SHEET

Muscle Activity in Multidirectional Instability of the Shoulder

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Investigators: J.M. Barden, R. Balyk, V.J. Raso, K. Bagnall, P.Zehr, M. Chan, M. Moreau

PURPOSE OF THE STUDY: To see if the muscles of people with unstable shoulders work differently than people with normal shoulders.

BACKGROUND: Shoulder muscles move and steady the shoulder. They must pull at the right time and with enough force to do this. Your body uses information about shoulder movement and position to control these muscles. It is thought that an inability to use this information may exist in people with unstable shoulders. Injury to the structures that provide this information may result in instability through abnormal muscle activity. We will examine how shoulder muscles work during three separate movements.

PROCEDURE: Five sets of three electrodes will be attached to muscles around your shoulder. Two sets of electrodes will be inserted into the skin using a small needle. You will be required to raise your arm forwards, upwards, backwards and to the side. These movements will be performed lying on your back, your side and from a seated position. A machine called a KinCom will control the movement direction and provide light resistance to your arm. We will show you how to do these movements in the correct position and at the proper speed. You will also be given verbal instructions about how to do this. You can practice to become familiar with the movements. Each movement will be performed ten times for each arm. The test procedure will take approximately three hours to complete. All subjects will be requested to perform only one test.

RISKS/BENEFITS: Some discomfort may occur with inserting the needle for the wire electrodes. Once the needle is removed, no discomfort should exist when performing the movements. All electrodes are sterile. No danger of electrical shock exists with these tests. These tests will help us to understand shoulder problems better and may suggest new treatments for this type of problem.

CONFIDENTIALITY/FREEDOM TO WITHDRAW: Data will be coded (no information identifying you will be used). It will be kept for at least 7 years following the end of the study. Data will be stored electronically (CD-ROM) and locked in a secure filing cabinet. The data will only be accessible to immediate members of the research group. Your decision to help in this study is voluntary. You are free to withdraw at any time. This will not affect any treatment provided to you by your physician, the hospital or its staff. Data from this experiment may be used in future studies. Should this be the case, further approval from an ethics committee is required. All information will be confidential except when professional codes of ethics and or legislation require reporting.

If you have questions about the study please contact Mr. Jim Raso at 471-2262, ext. 2390. Please initial below to indicate that you have read and understand the information described on this page.

Initial of Research Participant _____

Initial of Investigator _____

CONSENT FORM

Muscle Activity in Multidirectional Instability of the Shoulder

Principal Investigator(s): **Dr. R. Balyk , Dept. of Surgery, 439-5858,** Fax: **433-1344**

Co-Investigator(s): **John M. Barden, M.Sc., Dept. of Surgery, 471-2262 ext. 2688**

Part 2 (to be completed by the research subject):

Do you understand that you have been asked to be in a research study?	Yes	No
Have you read and received a copy of the attached Information Sheet?	Yes	No
Do you understand the benefits and risks involved in taking part in this research study?	Yes	No
Have you had an opportunity to ask questions and discuss this study?	Yes	No
Do you understand that you are free to refuse to participate or withdraw from the study at any time? You do not have to give a reason and it will not affect your care.	Yes	No
Has the issue of confidentiality been explained to you? Do you understand who will have access to your records?	Yes	No
Do you want the investigator(s) to inform your family doctor that you are participating in this research study? If so, please provide your doctor's name: _____	Yes	No

This study was explained to me by: _____

I agree to take part in this study.

_____ _____ _____

Signature of Research Participant Date Witness

_____ _____

Printed Name Printed Name

I believe that the person signing this form understands what is involved in the study and voluntarily agrees to participate.

_____ _____

Signature of Investigator or Designee Date

THE INFORMATION SHEET MUST BE ATTACHED TO THIS CONSENT FORM AND A COPY GIVEN TO THE RESEARCH SUBJECT